Plantar Pressure Measures of Running and Cutting Movements on Third-Generation Artificial Turf and Natural Grass

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Declaration

This work is original and has not been

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Signed………Matthew Page………….

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Abstract

In dynamic team sports such as rugby, football and American football, non-uniform differences in Anterior Cruciate Ligament (ACL) and ankle sprain injury rates have been found between third generation artificial turf and natural grass surfaces (Dragoo & Braun, 2010; Fuller, Dick, Corlette & Schmalz, 2007b). The purpose of this study was to assess differences in plantar pressure measures, during a straight run, 45 ° and 90 ° cut, performed on each of these surfaces.

Eight male university rugby players completed three trials of each movement on both a third generation artificial turf and natural grass surface. Speed was controlled, and pressure insoles were used to collect peak pressure, peak force, pressure-time integral and relative load data under the Medial Heel, Lateral Heel, Midfoot, Medial, Central and Lateral Forefoot, Hallux and Lesser Toes.

No surface effect was found for any of the above variables. A significant movement effect was found, whereby cutting increased peak pressures, force and pressure-time integrals under the Medial Heel, Lateral Heel, Medial Forefoot, Central Forefoot and Hallux. Cutting reduced peak pressures, forces and pressure-time integrals under the Lateral Forefoot and Midfoot. Relative load data suggested a medial shift in loading underneath the foot during cutting compared to running; however larger increases in loading at the heel may have masked differences in loading at other foot regions. The results suggest that no surface is likely to increase ACL or ankle sprain risk. Further study is needed to establish the causes of differences in injuries sustained on these surfaces.

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Introduction/Literature Review:

Artificial Turf and Natural Grass Surfaces

The majority of team sports are traditionally played outdoors on natural grass surfaces. However, since the 1960's, artificial turf surfaces have been developed and installed in place of natural grass pitches, for numerous reasons. Whilst natural grass is regarded as the optimum surface for most team sports, in certain climatic conditions grass struggles to survive. Furthermore, natural grass cannot withstand high levels of usage and extreme weather conditions, such as heavy rainfall, extreme cold or heat (Stiles, Guisasola, James & Dixon, 2011). Not only are artificial turf pitches superior in their durability, climatic resistance and drainage systems, but they are arguably more economical and cost effective (IRB, 2012; The Football League 2012). Artificial turf surfaces reduce the need for separate training facilities, artificial lighting to keep natural grass growing, natural grass maintenance, and match day losses, resulting from weather conditions making grass pitches unusable. There are pro-turf groups who believe that the reduced amount, but greater expense, of the more specific maintenance that artificial surfaces require makes them more expensive than grass surfaces (Turfgrass Resource Centre, Unknown), however it is generally believed that artificial turf creates a more cost-effective and economical option for most sports teams.

The development of artificial turf surfaces has led to some sports, such as hockey, adopting an artificial surface as their sole playing surface, superseding natural grass. Other sports, such as Football, American football, Rugby Union and Rugby League, have accepted artificial turf as an acceptable alternative sports surface to natural grass. In 2004 FIFA first included artificial turf in the laws of the game, with numerous teams adopting the surface (The

Football League, 2012). This is particularly the case in Scandinavia and Russia, where climatic conditions make grass difficult to grow and maintain. In December 2010 the IRB launched the One Turf program, in order to redefine artificial turf as a suitable playing surface for rugby (International Rugby Board, 2012). The One Turf program also aimed to create standards, alongside FIFA, in order to develop and maintain quality artificial surfaces. Currently, in the United Kingdom, both Rugby Union and Rugby League have one premiership and one Super League team using an artificial turf pitch, whilst 11 other rugby clubs share an artificial pitch with Football league clubs.

Artificial turf surfaces need to mimic the natural grass pitches that most sports are traditionally played on. The first generation artificial turf surfaces were characterised by short, high density fibres and a hard base layer (Levy, Skovron & Agel, 1990). The increase in shoe-surface traction and hardness of these surfaces led to an increase in impact, friction and cleat-catch-related injuries (Soligard, Bahr & Andersen, 2010). Second-generation artificial turf surfaces were modified, through the inclusion of less dense fibres and a granular infill, typically sand. This surface more closely recreated ball roll characteristics of natural grass; however differences in hardness and abrasion characteristics once again caused an increase in impact and friction-related injuries. Third-generation artificial turf surfaces have since been developed, and it is this generation that has been accepted by most team sport National Governing Bodies as an alternative to natural grass. Third-generation artificial surfaces are comprise of more grass-like fibres, made of polyethylene and lubricated with silicone, to reduce abrasion (Turfgrass Resource Centre, Unknown). A shockpad layer has also been added, in order to reduce impact forces and impact-related injuries. Shockpad layers either consist of a foam layer or a shredded rubber crumb infill.

It is though these developments have made third-generation artificial surfaces more grass-like. In terms of physiological and performance characteristics, artificial turf and natural grass have been shown to perform similarly. No difference in heart rate responses during incremental running or match simulation protocols have been found between the two surfaces (Hughes, Birdsey, Meyers et al., 2013; Michele, Renzo, Ammazzalorso & Merni, 2009). No difference in sprint time has also been found between the two surfaces (Gains, Swedenhjlem, Mayhew, Bird & Houser, 2010), however significantly shorter times have been recorded for completion of an agility course on artificial turf (Gains et al., 2010; Hughes et al., 2013). These differences have been attributed to the greater traction properties of artificial turf surfaces. Whilst Hughes et al. (2013) also recorded no differences in blood lactate responses between artificial turf and natural grass during a football match simulation protocol, Michele et al. (2009) recorded significantly lower running speeds needed to reach the onset of blood lactate accumulation, on artificial turf surfaces compared to natural grass. This suggests that running on artificial turf requires more energy, and is likely to invoke muscular fatigue sooner.

Epidemiology of Team Sports, Artificial Turf and Natural Grass Injuries

As well as differences in the performance of the sport surface, the injury potential of the sports surface needs to be considered. In sports such as rugby and American Football, players frequently collide with the ground, in jumping, running, cutting and driving movements. The intermittent and evasive nature of such sports means that, during training and match play, large amounts of stress are put on the lower extremity joints. It is well recognised that increases in the hardness and traction of sports surfaces increases the likelihood of injury (Stiles, James, Dixon & Guisasola, 2009; Kordi, Hemmati, Heidarian & Ziaee, 2011). Greater stress on the shock absorbing mechanisms of the lower body, and excessive joint torques created through the fixation of the foot in the ground during a rotational movement, are thought to be the two main causes of injury for surfaces with greater hardness and traction properties.

The lower limbs are the most commonly injured anatomical regions in rugby, comprising 42-59 % of all injuries at the elite level (Pearce, Brookes, Kemp & Calder, 2011). In a study of 899 premiership rugby union players over 4 years, injuries to the foot were accountable for 4 % of all injuries and 5 % of all injury-related absences (Pearce et al., 2011). Sprains to the Hallux accounted for 11 % of all foot injuries and 10 % of total foot injury absences, other foot ligament sprains accounted for 26 % of all foot injuries and 17 % of total foot injury absences, whilst 5th Metatarsal and Navicular stress fractures accounted for 2 % of all foot injuries, 3 and 11 % of total foot injury absences respectively (Pearce et al., 2011). It was hypothesised by Pearce et al. (2011) that higher ground reaction forces experienced in dynamic rugby movements such as cutting, jumping and landing may be responsible for stress fractures, whilst foot sprains are commonly caused by fixation of the foot within the sports surface. The significant absences caused by injuries to the foot, which totalled 3542 days for 147 injuries (Pearce et al., 2011), highlights the need for a sports surface that can reduce this injury potential. Given that the causes of foot injuries in the study by Pearce et al. (2011) correspond with the characteristics of artificial turf pitches, it could be assumed that artificial turf pitches are more likely to cause lower limb sprain and stress fracture-type injuries. Any increased injury potential of one surface over another may have dramatic consequences in terms of its acceptance as a suitable sports surface for dynamic team sports such as rugby. Because of this, a recent surge in research into injury differences between artificial turf and natural grass surfaces has occurred, although primarily focused on football.

Three studies have focused on injury-differences between third-generation artificial turf and natural grass in American Football (Dragoo & Braun, 2010; Dragoo, Braun & Harris, 2012; Meyers & Barnhill, 2004). Eight studies have looked at injury rates in football played on third-generation artificial turf and natural grass (Hagglund, Walder, Zwerver & Ekstrand, 2011; Ekstrand, Hagglund & Fuller, 2011; Bjornboe, Bahr & Andersen, 2010; Ekstrand, Timpka & Hagglund, 2006; Fuller, Dick, Corlette & Schmalz, 2007a, 2007b; Steffen, Andersen & Bahr, 2007; Soligard, et al., 2010). Only one study has compared injury rates between third-generation artificial turf and natural grass in rugby (Fuller, Clarke & Molloy, 2010).

The non-uniform methods of reporting injury data make it difficult to collate results from different studies. However some main trends can be observed between the two surfaces. In American Football, a significantly higher number of acute, non-contact or running-related lower body injuries were sustained on artificial turf compared to natural grass (Dragoo & Braun, 2010; Meyers & Barnhill, 2004). This was reported as 44.29 % of all injuries, compared to 36.12 %, by Dragoo and Braun (2010) and an injury incidence rate of 1.0 vs. 0.3 by Meyers and Barnhill (2004) for artificial turf and natural grass pitches respectively. Meyers and Barnhill (2004) also reported a significantly higher number of muscle-tendon overload injuries (1.1 vs. 0.3), muscle strains

(2.1 vs. 1.1), overall muscle injuries (5.5 vs. 4.0), and injuries with a 7-21 day rehabilitation period (1.9 vs. 1.3) on third-generation artificial turf versus natural grass respectively. A higher number of concussion injuries (1.8 vs. 0.7), ligament tears (1.0 vs. 0.5) and Anterior Cruciate Ligament (ACL) injuries (1.0 vs. 0.4) were reported on grass compared to third-generation artificial turf (Meyers & Barnhill, 2004). Conversely, Dragoo and Braun (2010) reported a higher number of ACL injuries on artificial turf compared to natural grass (1.73 ACL injuries per 10000 athlete exposures vs. 1.39/10000 athletes exposures). This conflict may be due to Dragoo and Braun (2010) reporting the number of athlete exposures, as opposed to the duration of play. Athlete exposure does not give an indication of the amount of time on the field; therefore it cannot be assumed that the injuries reported in this study occurred over the same, or a relative, amount of time.

In football, no difference in the overall incidence of injuries has been recorded for males or females, during training or matches, between third-generation artificial turf and natural grass (Steffen et al., 2007; Ekstrand et al., 2011; Ekstrand et al., 2006; Bjornboe et al., 2010; Soligard et al., 2010; Fuller et al., 2007a, 2007b). For females, a higher number of ankle and knee injuries on artificial turf compared to grass during matches has been reported (Steffen et al., 2007), with values of 1.4 ankle and 1.7 knee injuries occurring on artificial turf for every 1 on grass. However Fuller et al. (2007a) reported a greater number of ankle injuries for females during matches on grass compared to artificial turf, with 4.21 ankle injuries per 1000 playing hours on grass versus 3.0 per 1000 on artificial turf. This conflict in results is likely to be influenced by the small number of injuries recorded in these studies, so it is possible that type II errors have been made.

Amongst elite males a significantly higher number of ankle injuries have been found on third generation artificial turf (Fuller et al., 2007b; Ekstrand et al., 2011) Fuller et al. (2007b) reported 0.83 ankle injuries per 1000 training hours on turf compared to 0.58/1000 on grass. Ekstrand et al. (2011) reported rates of ankle injury at 4.8/1000 match play hours on turf compared to 3.53/1000 on grass. Other studies have reported no difference in ankle injuries between artificial turf and natural grass in males during training (Ekstrand et al., 2006; Bjornboe et al., 2010; Ekstrand et al., 2011) or matches (Fuller et al., 2007a; Ekstrand et al., 2006). However trends towards a greater number of ankle sprains on artificial turf have been recorded in these studies, with Ekstrand et al. (2006) recording nearly twice the incidence of ankle sprains of artificial turf compared to grass (4.83/1000 playing hours vs. 2.66/1000 playing hours). Contrary to these results, Soligard et al. (2010) recorded nearly twice the number of ankle sprain injuries on grass compared to artificial turf (8.4/1000 playing hours vs. 4.3/1000 playing hours).

Contrary to the studies in American Football, no effect of surface has been found for the number of knee or ACL injuries occurring during training or matches, for males and females, in football (Fuller et al., 2007b; Ekstrand et al., 2011). Trends towards a higher number of ACL injuries on third generation artificial turf have been reported (Bjornboe et al., 2010; Fuller, 2007a), as well as a conflicting trend towards a greater number of ACL injuries on grass (Soligard et al., 2010), but no significant differences have been recorded. Very few consistent results exist between studies assessing the effect of surface on injury risk in football. Surface does not seem to have an effect on the overall injury risk, or severity of injury. The surface does seem to affect certain kinds of injury surrounding the knee and ankle complex, but not in a uniform manner.

This is likely to be due to the lack of conformity in the type of surface, as both grass and artificial surfaces are manufactured and maintained at different standards. The small number of specific injuries, such as ACL injuries, reported in these studies is also likely to be responsible for a number of type II errors, making conclusions on the effect of surface on injuries in football difficult to draw.

The single study to investigate the effects of surface on injury potential in rugby union also reported the greatest injury occurrence in the lower limbs during training (77.8 % on artificial turf, 68.1 % on grass) and matches (57.7 % on artificial turf, 39.3 % on grass) (Fuller, et al., 2010). No difference in the overall incidence of injuries was found between surfaces, with incidences of 3.0/1000 playing hours for artificial turf and 2.3/1000 playing hours on grass. A non-significant increase in ankle and knee ligament injuries was reported on artificial turf compared to natural grass. This again will likely be due to the small numbers of these injuries, with 5 ankle ligament injuries sustained on artificial turf, compared to one on grass, and 11 knee ligament injuries on artificial turf compared to 3 on grass. The incidence rate of ACL injuries was 3.7/1000 playing hours on artificial turf and 1/1000 playing hours on grass.

In a number of team sports, surface (grass or third-generation artificial turf) seems to have no effect on the overall number of injuries or severity of injuries sustained. An increased number of ACL injuries have been reported on artificial turf in American footballers, and a trend towards this has been observed in rugby. Surface also seems to have an effect on ankle ligament injuries, although not a uniform effect. This is likely to be affected by the small number of injuries recorded in these studies, so no detailed conclusions can be made.

Aetiology of Anterior Cruciate Ligament (ACL) and Lateral Ankle Sprain Injuries

Both ACL and ankle ligament injuries require a lengthy rehabilitation period and place a heavy financial strain on national health services. Whilst figures are not available for the United Kingdom, treatment of ACL injuries in the USA costs around 3 billion dollars annually (Frobell, Roos, Roos, Ranstam & Lohmander, 2010). Ankle ligament sprains have been reported in English premiership football to have a recurrence rate of 9 % (Woods, Hawkins, Hulse & Hodson, 2003), whilst ACL injured patients are up to 10 times more likely to develop osteoarthritis (Hewett, Ford & Meyer, 2006).

The ACL connects the distal portion of the Femur to the Tibial plateau, and is placed under strain during internal rotation of the shank, or anterior movement of the Tibia (Drakos, Hillstrom, Voos et al., 2010; Dempsey, Lloyd, Elliott et al., 2007). The most common ankle ligament injuries are to the lateral ligament complex. This consists of the Anterior Talofibular, Calcaneofibular, and Posterior Talofibular ligaments. The calcaneofibular ligament is the primary stabiliser of the Subtalar joint, and the lateral ligament complex works to control and limit the maximal range of ankle inversion and dorsiflexion (Kliepool & Blankevoort, 2010). Common ankle lateral ligament complex injuries are inversion sprains, and are caused by excessive ankle inversion or supination placing excess strain on the ligaments on the lateral portion of the Subtalar joint (Kliepool & Blankevoort, 2010). A common sporting movement that typically involves internal rotation of the shank and ankle inversion is a cutting movement. Cutting movements, whereby the athlete produces a rapid change of direction with the plant foot in contact with the ground, are reported to be responsible for most ankle and ACL ligament injuries during team sports (Kristianslund, Bahr & Krosshaug, 2011; Dempsey et al., 2007; Vanrenterghem, Venables, Pataky & Robinson, 2012; Smeets, Jacobs, Hertogs et al., 2012; Dayakidis & Boudolos, 2006; Donnelly, Lloyd, Elliott & Reinbolt, 2012; Ford, Meyer, Toms & Hewett, 2005; Gehring, Wissler, Mornieuz, & Gollhoffer, 2013; Bahr & Krosshaug, 2005). Between two-thirds and three-quarters of all ACL injuries are non-contact injuries, sustained during a cutting manoeuver (Donnelly et al., 2012; Smeets et al., 2012; Kobayashi, Kanamura, Koshida et al., 2010; Boden, Sheehan, Torg & Hewett, 2010), however the underlying mechanisms causing these injuries are multifactorial.

The ankle joint has a maximal inversion-eversion range of motion of around 17-23 °, controlled by the lateral ankle ligament complex (Kliepool & Blankevoort, 2010). In the functional range of motion of the ankle joint, the lateral ankle ligaments are slack, with zero or small loading of the Anterior Talofibular and Calcaneofibular ligaments between 10 ° of dorsiflexion and 20 ° of plantarflexion. In two studies that have captured kinematic data on accidental ankle inversion sprains when performing a cutting movement, injuries have been characterised by high ankle inversion and inversion velocity (Kristianslund et al., 2011; Gehring et al., 2013). Kristianslund et al. (2011) recorded 23 ° of ankle inversion in the injured trial, with a maximum ankle inversion velocity of 559 °/s. This was compared to an ankle inversion velocity of 166 and 221 °/s in control subjects. Gehring et al. (2013) recorded an ankle inversion velocity of 1290 °/s in the injured trial, but also found that the injured trial was initiated by the proximal segments, with a delayed hip flexion invoking a more extended knee at touchdown, and resulting in a steep and rapid heel strike. Neither of these studies modeled the Subtalar or Talocrural joint separately, so the individual roles of joints within the ankle have not been quantified. However it appears a combination of high ankle inversion and inversion velocity increases the likelihood of a lateral ankle sprain. Greater ankle inversion is suggested to be caused by a medially deviated centre of mass at foot contact (Fong, Chan, Mok, Yung & Chan, 2009). A medial deviation of the centre of mass with respect to the Subtalar joint would create a greater moment arm along the joint axis. This places greater strain on the lateral ankle ligaments to counter the external inversion moment, and is the probable cause for lateral ankle ligament injury.

Injuries to the ACL ligament are suggested to be caused by: high knee valgus moments; an extended knee at heel contact; impingement of the ACL against the intercondyle notch; internal rotation of the knee; hamstring-quadriceps imbalance; weight bearing; a posterior shift of the centre of mass behind the base of support; and an anterior shift of the tibia relative to the femur (Vanrenterghem et al., 2012; Boden et al., 2010; Ebben, Fauth, Petushek et al., 2010; Jorrakate Vachalathiti, Vongsirinavarat, & Sasimontonkul, 2011; Hewett, Ford, Hoogenboom & Myer, 2010; Bahr & Krosshaug, 2005; Dempsey et al., 2007; Kobayashi et al., 2010). Some of these causes hold more merit than others. Ebben et al. (2010) and Hansen, Padua, Blackburn, Prentice and Hirth (2008) deduce that, with the Quadriceps being the primary producer of anterior knee force with the knee at full extension, excessive Quadriceps force will strain the ACL ligament, which aims to hold the tibia posteriorly. However Boden et al. (2010) argue that the angle of the patella tendon at knee full extension is shallow, and therefore the Quadriceps create a compressive tibia-femoral force at the angle, which will not anteriorly displace the Tibia.

Little credit has also been given to the theory that notch impingement is the cause of ACL injuries. A smaller intercondyle notch, which is suggested to be a genetic trait, would suggest a smaller ACL, more susceptible to strain under a smaller range of motion. No studies however have looked to assess the prevalence of smaller epidcondyle notches in different populations, so this claim cannot be supported.

The most supported cause of ACL injury is that of an extended knee at heel contact, weight bearing, knee internal rotation and valgus, and a posterior shift in the centre of mass (Boden et al., 2010; Alentorn-Geli, Myer, Silvers et al., 2009). This combination of factors is believed to increase the likelihood of an anterior Tibia shift. As the ACL holds the Tibia posteriorly, it is the anterior shift of the Tibia, caused by these factors, which ultimately leads to ACL rupture or tear. Excessive knee extension at heel contact suggests a more flexed hip position, and therefore a more posterior centre of mass. This also means that the Tibial plateau is aligned more vertically (Boden et al., 2010). The vertical alignment of the Tibial plateau puts compressive forces on the posterior section of this slope, promoting a posterior shift of the Femoral condyle, and an anterior shift of the Tibia (Boden et al., 2010). The amount of vertical force experienced during weight-bearing activities causes this displacement of the Tibia from the Femur, and puts excessive and potentially injurious strain on the ACL. Excessive knee valgus during this movement will also add to the strain on the ACL, by compressing the medial knee compartment and allowing anterior Tibial shift on the lateral portion of the knee (Boden et al., 2010).

During cutting tasks, peak knee Valgus has been found to be two times that of normal running (Donnelly et al., 2012). Postural adjustments have been found to reduce peak knee valgus moment, and therefore ACL strain, from 106.1 to 61.9 Nm, through repositioning the centre of mass (Donnelly et al., 2012). Repositioning of the centre of mass towards the centre of the knee joint serves to reduce compression on the medial knee compartment; reducing joint laxity on the lateral side and preventing an anterior Tibia shift. To demonstrate this, Dempsey et al. (2007) recorded higher peak knee valgus values for a wide foot position during a 45 ° cut (0.79 Nm.Kg) compared to normal foot position (0.45 Nm.Kg) and a medial foot placement (0.51 Nm.Kg). Furthermore trunk lean towards the plant foot, increasing medial compression further, produced peak knee valgus values of 0.65 Nm.Kg. Besier, Lloyd, Ackland and Cochrane (2001) found that unanticipated cutting served to reduce the time taken to adequately make postural adjustments, and therefore increased knee valgus loading. Besier et al. (2001) recorded knee valgus moments 1.5 times greater than during anticipated cuts for unanticipated cuts at 60 °, with a 12 fold increase at 30 °. Knee internal rotation, a movement which serves to tighten the ACL, also increased during unanticipated cutting, by 49 % for 60 ° and 129 % for 30 ° cuts.

There are a multitude of non-surface-specific factors affecting both ACL and ankle ligament injuries. The significant financial and health costs of ligament injuries to the knee and ankle highlight the need to establish the propensity of artificial turf and natural grass pitches to cause injury in team sports, which incorporate a large amount of dynamic movements such as cutting. Whilst there are many highlighted benefits of artificial turf surfaces, the increase in use of these surfaces as a team sports pitch would depend upon the financial and injury cost they invoke. Likewise, a decrease in the propensity of artificial turf surfaces to cause injury would likely increase their popularity and use, given their many economic benefits.

Mechanical Properties of Sport Surfaces

Hardness and traction of sports surfaces are widely regarded as being related to injury (Stiles et al., 2009). Hardness in particular is likely to affect ACL and ankle sprain injuries, as increased ground hardness is likely to accentuate a steep heel strike, which is a potential kinematic precursor for ACL and ankle inversion sprain injuries. In rugby, trends towards lower ground hardness, corresponding with a non-significant drop in lower body injuries, have been found as the season progresses on natural grass pitches (Orchard, 2002). However the exact relationship between ground conditions and injury risk is difficult to calculate, due to the multivariate nature of the weather. There is also a lack of objective measurement of ground conditions, making correlations with injuries hard to make. Only Orchard (2002) assessed ground hardness on natural grass pitches, using a penetrometer. Orchard (2002) found a non-significant decrease in ground hardness resulted in a significant drop in ACL risk.

The association between hardness of artificial turf surfaces and injury risk has not been assessed. This is likely to be because the mechanical properties of artificial surfaces are more uniform than natural grass pitches, due to the strict requirements enforced by International Governing Bodies such as FIFA and the IRB. Artificial turf surfaces must pass a series of lab and field test before being deemed acceptable to use for sports such as rugby and football. Mechanical tests are implemented for: shock absorbency; head injury criterion; vertical deformation; traction; slip resistance; abrasiveness; energy restitution;

vertical ball rebound; pile height; joint strength and durability; rotational resistance; linear friction; slope; and water permeability (Doran, 2010; International Rugby Board, 2012). It is unclear how acceptable levels for each of these mechanical variables have been defined. Recommendations for hardness of artificial turf surfaces have even been made on the statement that shock absorbency is a measure of comfort, not performance, and is not a safety test (Doran, 2010). With the inability to sufficiently absorb ground reaction forces in the bodies various tissues the cause of most chronic or overuse lower body injuries and fractures, the relevance of these safety measures in relation to the human system should be put to question.

Artificial turf surfaces are tested using mechanical equipment, such as Clegg Hammers and Advanced Artificial Athletes. The lack of human testing on artificial surfaces casts further doubt on the acceptability of supposed ‘safe’ values of its mechanical properties. Mechanical tests are simplifications of the human system, incapable of representing any time-dependant pressure or force factors under the foot (Stiles et al., 2009; Hennig 2011). Furthermore mechanical tests of hardness cannot reflect human multi energy-absorbing strategies or multi-planar forces and velocities (Saunders, Twomey & Otago, 2011; Kent, Crandall, Forman et al., 2012). Therefore mechanical tests of ground hardness are unable to represent the multi-planar forces experienced during dynamic sporting movements such as cutting, which are a frequent occurrence in team sports.

Saunders et al. (2011) assessed the relationship between human loading characteristics and Clegg Hammer measures. In drop jump landings, maximum vertical ground reaction force and vertical loading rate, as measured on humans using a force platform, correlated moderately (.654) on a surface with no shock absorbing layer. On a 15 mm and 50 mm shockpad layer, the two measures still correlated moderately in humans (.651 and .543 for 15 mm and 50 mm respectively). Correlations between Clegg hammer measures of ground hardness and maximum vertical ground reaction force varied between .001 and .069 for the three surfaces. Clegg hammer measures of vertical loading rate and human-measured vertical loading rate correlated between -.171 and .088. Therefore no relationship exists between mechanical tests of sports surfaces and human kinetic values, and mechanical tests cannot be advocated in the assessment or monitoring of the injury potential of artificial turf surfaces.

Human testing of ground hardness, and the effects of harder surfaces, has typically used force platforms to measure peak vertical impact forces and loading rates. Vertical impact loading rate is regarded as the best predictor of chronic or overuse lower body injuries, as it gives force measures as a function of time, so is more reflective of the demands placed on the lower body (Stiles & Dixon, 2007; Dixon, Collop & Batt, 2000). Stiles and Dixon (2007) found contradicting results when assessing four surfaces of different levels of hardness. The softer 15 mm thin foam and 45 mm thick foam surfaces recorded significantly lower vertical loading rates (362 BW/s and 335BW/s for the thin and thick foam respectively) than the acrylic (756 BW/s) and rubber (660 BW/s) surfaces during a tennis forehand foot plant. However peak vertical impact forces were higher on the softer surfaces (3.58 and 3.25 times body weight for the 15 and 45 mm foam surfaces respectively) than the harder acrylic (2.66 times body weight) and rubber surfaces (2.66 times body weight). Stiles and Dixon (2007) established that adaptations to foot strike patterns to attenuate larger ground reaction forces on harder surfaces may change kinetic impact values. An alternative and more sensitive measure of surface cushioning recommended by Stiles and Dixon (2007) was peak heel pressure. Peak heel pressure reduced from the hardest acrylic (34.8 $N.cm^2$), softer rubber (31.4 $N.cm^2$), then the thin foam (23.9 $N.cm^2$) and thick foam surfaces (21.6 $N.cm^2$). A more appropriate method of measuring the propensity of natural grass and artificial turf surfaces may therefore be measures of pressures underneath the foot, during dynamic sporting movements.

The Use of Pressure Insoles for Assessing Injury Risk

Pressure insoles are devices inserted into the shoes, to assess pressures applied to the whole plantar surface of the foot, at the foot-shoe interface (Tillman, Fiolkowski, Bauer, & Reisinger, 2002). This load is reflective of the load applied to the body, and is not inclusive of the load absorbed by the various cushioning properties of the shoe. Therefore pressure insoles may be a better predictor of injury potential than force platforms, which measures forces applied at the shoe-ground interface. Pressure insoles allow for simple data collection and are not limited, as force platforms are, to analysis of a single step (Dixon & McNally, 2008; Rouhani, Favre, Crevoisier, & Aminian, 2010). Pressure insoles also allow for data collection in a more ecologically valid setting. Lab studies typically use small, isolated samples of artificial turf and natural grass surfaces. These small samples are influenced by the boundaries they are installed in, which are not reflective of their environmental boundaries, and therefore it must be recognised that artificial turf and natural grass surfaces need to be installed in their natural environments in order to act correctly (Kent et al., 2012). The requirement for field testing on different surfaces necessitates the use of pressure insoles, given the difficulty and cost involved in the installation of force platforms in-situ.

Pressure insoles allow for the collection of numerous kinetic and kinematic variables. The numerous sensors that comprise each pressure insole can be used to break the foot down into smaller areas that represent anatomical regions of the foot, such as the heel or metatarsal heads. The pressure, contact time and contact area data automatically collected by pressure insoles software can then be used to calculate force values under each predefined area of the foot, as well as values of impulse, expressed in both force and pressure terms. Values of impulse allow for a more valid interpretation of the demands placed on different foot regions, taking into account both the load applied to the area, and the time over which the load is applied. Pedar pressure insoles, a common and popular brand of pressure insole, have been found to be highly repeatable in a number of these kinetic and kinematic variables. Putti, Arnold, Cochrane and Abboud (2007) recorded a coefficient of repeatability of less than 10 % in 93.4 % of peak pressure, pressure-time integral (impulse), contact area, contact time and force-time integral (impulse) variables over 10 areas of the foot during walking. Over the heel, lateral midfoot, 1^{st} metatarsal head, 2^{nd} metatarsal head, 3^{rd} metatarsal head, 4^{th} metatarsal head, 5^{th} metatarsal head, hallux, 2^{nd} toe and 3^{rd} to 5^{th} toes, only the 2^{nd} left toe recorded a coefficient of repeatability of over 10 %, for peak pressure, pressure-time integral and force-time integral values. Pressure-time integral also had a greater coefficient of repeatability in the 2^{nd} toe and 3^{rd} to 5^{th} toes, however these values were all deemed acceptable, within 15.3 %. Ramanthan, Kiran, Arnold, Wang, and Abboud (2010) also recorded highly repeatable values of pressure, force, contact area, contact time and force-time integral under the heel, midfoot, 4^{th}, 3^{rd} and 2^{nd}

metatarsal heads during walking. Acceptable repeatability levels for these variables were found under the 1st metatarsal heads, hallux, 3rd to 5th toes and 2nd toe. The 2nd toe again was found to be the least repeatable anatomical region; however these values were still acceptable.

Pressure insoles only allow measure vertical forces, no shear forces are calculated. Whilst algorithms have been developed to calculate the complete ground reaction forces from pressure insoles data, to within 5 % of force platform values, during running, cutting and jumping (Fong, Chan, Hong et al., 2008b) vertical forces at the foot-shoe and shoe-ground interface are most commonly associated with injury. Pressure insoles have been shown to consistently underestimate values of force when compared to a force platform. Barnett, Cunningham and West (2001) found consistent measures of contact time for pressure insoles versus a force platform during walking (657 vs. 668 ms for force platform vs. pressure insoles). Despite the underestimation of force, by pressure insoles, in the impact peak (1.25 vs. 1.07 times body weight, force platform vs. pressure insoles) and propulsive peak (1.14 vs. 1.05 times body weight, force platform vs. pressure insoles), stable timing parameters allowed for an accurate comparison of force-time integral (55 vs. 53 Ns, force platform vs. pressure insoles). Lower force values would be expected from pressure insoles, firstly because lower sample rates suggest high frequency data loss, and secondly, because force dissipated by the shoe will not be present at the foot-shoe interface (Barnett et al., 2001). So whilst algorithms have also been made to increase the validity of vertical ground reaction forces from pressure insoles (Cordero, Coopman & Van Der Helm, 2004), the accurate calculation of unadjusted pressure-time and force-time variables and expected lower force values make pressure insoles an accurate and repeatable method for assessing injury risk in the field, during different dynamic tasks.

An under-reported use of pressure insoles is their ability to detect changes in lower body kinematics. Through the insertion of orthotic devices in the shoe, Dixon and McNally (2008) found that reduced pressure on the medial side of the foot corresponded with increased rearfoot inversion, influenced by the orthotic device. Therefore greater inversion of the foot, a condition that, during dynamic activities, is a possible premise for ankle inversion or ACL injuries, manifests itself as a reduced pressure on the medial side of the foot. Erhart, Mundermann, Mundermann, and Andriacchi (2008) used pressure mats to predict knee joint moments in slow, normal and fast walking with a 4 ° and 8 ° wedged insole. In 13 of 15 participants, Erhart et al. (2008) reported a negative correlation between peak knee adduction moment and the ratio of maximum medial to lateral heel pressure. Two of the 15 participants recorded positive correlations between peak knee adduction moment and maximum medial to lateral heel pressure ratio, with correlation values between .004 and .8222 over the range of speeds. In most participants therefore, higher medial pressure was associated with greater foot eversion. Whilst not all participants followed this trend, it can be concluded that changes in the adduction moment or knee valgus, can be predicted with the use of pressure devices, but the magnitude of these changes cannot. Pressure insoles therefore can be used to assess foot in-eversion position and differences in knee adduction moment, increasing their appeal as a field tool for assessing ACL and ankle sprain injury risk.

It could be argued that pressure insoles, whilst allowing for more ecologically valid data collection, do not provide values for multi-planar lower body kinematic and so are not as useful as an integrated force platform/3D camera system. However, during dynamic sporting movements such as cutting, assessing lower body kinematics using 3D video and skin-mounted markers has been found to contain high levels of error. A comparison between skin-mounted markers and bone pins has shown up to 13.1 ° rotational error and 16.1 mm translational error in knee kinematics during cutting (Benoit, Ramsey, Lamontagne, et al., 2006). The highest levels of rotational error were found in the frontal plane, with 6.7 ° rotational error at heel contact and 5.9 ° at midstance. Sagittal and transverse plane rotations were found to contain 6.7 ° and 5.4 ° of error at heel contact and 4 ° and 5.4 ° of error at midstance. Translational error was highest in the medio-lateral direction at heel contact (7.3 mm), followed by the proximal-distal (6.3 mm) and anterior-posterior (5.6 mm) directions.

As knee kinematics are associated with ACL injuries, it is highly possible that differences in lower body kinematics between surfaces such as artificial turf and natural grass during cutting movements may fall into these boundaries for error associated with skin movement. For example Jones, Kerwin, Irwin, and Nokes (2009) reported differences in knee kinematics when landing on artificial turf and natural grass of 13 % in the medio-lateral direction. As up to 6.7 ° of rotational error has been found in the frontal plane, for the 13 % difference in frontal plane knee kinematics between turf and grass to fall outside the 6.7 ° margin of error, frontal plane knee range of motion would have to be around 52 °. This is well outside the frontal plane knee range of motion, and shows that assessing injury risk through the interpretation of lower body kinematics is wrought with error using skin-mounted markers. This perhaps advocates the use of pressure insoles in a field-test environment, over lab based turf studies, as pressure values can accurately be interpreted to infer about injury potential, and changes in knee and foot positions can be predicted.

Pressure insoles have previously been used to assess the injury-potential of popular running surfaces of differing levels of hardness. Tillman et al. (2002) used pressure insoles to calculate measures of force, contact time and impulse when running on asphalt, synthetic rubber, concrete and grass surfaces. Tillman et al. (2002) reported no difference between the four surfaces in total foot peak force measures (asphalt 1.38 body weight [BW], synthetic rubber 1.36 BW, concrete 1.36 BW, grass 1.46 BW), contact time (2.17.7, 216.2, 226.9 and 212.9 ms for asphalt, synthetic rubber, concrete and grass surfaces respectively) or impulse (132.6, 132.6, 136.3 and 134.5 Ns for asphalt, synthetic rubber, concrete and grass surfaces respectively). Measures of pressure take into account force over the area it is applied, and with pressure insoles more sensitive to measures of pressure than force, perhaps the small differences observed in this study would prove significant if measures of pressure were made. Furthermore Tillman et al. (2002) used a four region mask to break the foot down into front and back medial and lateral sides. A more anatomically accurate breakdown of the foot region may have found differences between the four surfaces, in terms of loading under specific areas of the foot, which would have been more useful in terms of assessing injury-potential of different surfaces.

Tessutti, Ribiero, Trombini-Souza, and Sacco (2012), when assessing the same four running surfaces, found significantly lower peak pressures when running on grass, under the medial rearfoot (276.1 Kpa vs. 306.4, 304.5 and 308.2 Kpa for grass vs. asphalt, concrete and synthetic rubber). Grass also invoked significantly lower peak pressures under the central rearfoot (299.5 Kpa vs. 347.7, 348.9 and 336.3 Kpa for grass vs. asphalt, concrete and synthetic rubber), lateral rearfoot (283 Kpa vs. 336.8, 337 and 339.5 Kpa for grass vs. asphalt, concrete and synthetic rubber), medial forefoot (337.7 Kpa vs. 361.9, 362.7 and 354.5 Kpa for grass vs. asphalt, concrete and synthetic rubber) and lateral forefoot (214.5 Kpa vs. 244.5, 242.3 and 242.5 Kpa for grass vs. asphalt, concrete and synthetic rubber). Tessutti et al. (2012) concluded that running on grass is less likely to cause chronic overuse injuries, as well as ACL and ankle sprain injuries, through placing less strain on anatomical areas of the foot. Running on grass also invoked a more neutral rearfoot position than running on asphalt, rubber or concrete. In fact the greater lateral pressures on these surfaces suggests that running on asphalt, rubber or concrete causes a more inverted foot position. Inversion of the foot is involved in impact attenuation (Low, 2010), so on harder surfaces this is expected, however this also means running on these surfaces predisposes runners to ankle inversion traumas and ACL injuries.

This lateral shift in plantar pressures on harder surfaces has been reported in other studies assessing running on concrete, rubber, grass and asphalt. Higher peak pressures have been reported on concrete compared to grass, under the total foot (451.8 vs. 401.7 Kpa for concrete vs. grass), lateral midfoot (175.3 vs. 148 Kpa), central forefoot (366.3 vs 336.8 Kpa) and lateral forefoot (290.2 vs. 257.9 Kpa) (Wang, Hong, Li & Zhou, 2012). Tessutti, Trombini-Souza, Ribiero, Nunes, and Sacco (2010) recorded significantly higher pressures, when running on asphalt compared to grass, under the central rearfoot (303.8 vs. 342.2 Kpa for grass vs. asphalt), lateral rearfoot (312.7 vs. 350.9 for grass vs. asphalt) and lateral forefoot (221.4 vs. 245.3 Kpa for grass vs. asphalt). Hong, Wang, Li and Zhou (2012) also reported lower peak pressures when running on grass compared to concrete, under the central forefoot (369.7 vs. 331.5 Kpa for concrete vs. grass) and lateral forefoot (300.3 vs. 247.6 Kpa for concrete vs. grass). It may be expected that harder surfaces invoke higher peak pressures under the foot, due to the greater magnitude of the vertical ground reaction force. Therefore it is difficult to decipher, through reporting raw pressure values, whether differences in peak pressures between surfaces are due to differences in foot position or magnitude of the ground reaction force. Reporting the relative load, as the percentage of load applied to each foot region, allows us to establish the magnitude of pressure applied to each foot region relative to the total force applied to the foot. Any changes in relative load therefore signify changes in foot position or loading of the foot. Reporting the relative load beside peak pressure data during running on different surfaces will allow for a better understanding of how the surface affects foot position and loading, regardless of the magnitude of the vertical ground reaction force.

Pressure insoles have also been used to assess plantar pressures during a range of dynamic sporting movements on one surface. Wong, Chamari, Mao, Wisloff, and Hong (2006) found that a 45 ° cut on third generation artificial turf induced significantly higher pressures than straight running, under the hallux (492 vs. 383 Kpa for cut vs. run), medial forefoot (550 vs. 367 Kpa for cut vs. run), medial heel (441 vs. 238 for cut vs. run), and lateral heel (401 vs. 247 for cut vs. run). A significantly higher pressure-time integral, a measure of impulse that takes into account the area-specific load under the foot as a function of time, was also found for the cut under the medial forefoot (106 vs. 47 Kpa.s for cut vs. run), medial heel (43 vs. 19 Kpa.s for cut vs. run) and lateral heel (40 vs. 20 Kpa.s for cut vs. run). Running caused significantly higher peak pressures (178 vs. 79 Kpa for run vs. cut) and pressure-time integrals (24 vs. 13 Kpa.s for run vs. cut) under the lateral forefoot. Again relative load has not been calculated to determine changes in plantar loading, however differences in pressure-time integral suggest that cutting induces greater stress on the medial side of the foot. This would suggest that cutting is less likely to cause inversion-related injuries, contrary to the epidemiological evidence. However increased pressure on the medial side of the foot suggests greater hip abduction and increased loading on the medial knee compartment. Increased loading on the medial knee compartment may cause valgus collapse and anterior tibial displacement, so may be the cause of ACL injuries during cutting in this data set.

Eils, Streyl, Linnenbecker et al. (2004) also found higher medial plantar pressures during cutting than in running. Performing a cut on a grass and red cinder surface caused greater peak pressures under the medial heel (655 vs. 298 Kpa for cut vs. run), lateral heel (489 vs. 294 Kpa for cut vs. run) and hallux (487 vs. 348 Kpa for cut vs. run). Furthermore, increased relative load was found under the medial heel (15.9 vs. 7.8 % for cut vs. run), lateral heel (11.3 vs. 8.8 % for cut vs. run) and hallux (13.3 vs. 9.6 % for cut vs. run), highlighting a definite medial shift in loading under the foot.

Pressure analysis has also been utilised to assess differences in plantar loading during tennis-specific movements (Girard, Eicher, Fourchet, Micallef, & Millet, 2007; Girard, Eicher, Micallef, & Millet, 2010), assessing gender differences in cutting (Sims, Hardaker & Queen, 2008), and the effect of foot arch height on plantar loading during running (Chuckpaiwong, Nunley, Mall &

Queen, 2008). The effect of footwear on plantar loading has also been assessed. Shoes with better cushioning properties have been shown to reduce pressures on the lateral side of the foot. Queen, Abbey, Wiegerinck, Yoder, and Nunley (2010) reported greater total foot force in racing flats compared to training shoes during running (2.8 vs. 2.5 BW for racing flats vs. training shoe), as well as higher lateral forefoot force (.429 vs .371 BW for racing flats vs. training shoe). Wiegerinck, Boyd, Yoder et al. (2009) also recorded higher total foot peak force for racing flats versus training shoes during running (2.74 vs. 2.49 BW), as well as greater total foot peak pressure (446.56 vs. 407.26 Kpa). Higher peak pressures were also recorded for racing flats under the lateral midfoot (201.39 vs. 184.19 BW for racing flats vs. training shoe), medial forefoot (358.83 vs. 323.76 BW for racing flats vs. training shoe), mid forefoot (336.87 vs. 311.02 BW for racing flats vs. training shoe), lateral forefoot (293.15 vs. 255.58 BW for racing flats vs. training shoe), hallux (383.61 vs. 331.83 BW for racing flats vs. training shoe) and lesser toes (330.72 vs. 301.93 BW for racing flats vs. training shoe). The increase in pressure loading under most areas of the foot here however highlights the need for analysis of relative load, to decipher whether shoes with lower cushioning properties affect loading pattern and are likely to increase ACL or ankle inversion injury risk on artificial turf or grass.

In comparisons of rugby and football footwear, astroturf boots have been found to allow for the greatest foot contact area during cutting on artificial turf (.903 NICA), compared to bladed boots (.903 NICA), 12-stud firm ground boots (.889 NICA) and 25-stud hard ground boots (.906 NICA), thereby increasing the area over which the ground reaction force can be dissipated (Queen, Charnock, Garrett et al., 2008). Astroturf boots therefore invoked the lowest total foot peak pressure (623.92 Kpa vs. 736.81, 758.57 and 755.41 Kpa for astroturf vs. bladed, firm ground, hard ground boots), as well as medial forefoot force (.518 BW vs. .552, .566 and .554 for astroturf vs. bladed, firm and hard ground boots) and medial forefoot force-time integral (86.04 Bw.s vs. 90.88, 93.87 and 90.61 Bw.s for astroturf vs. bladed, firm ground and hard ground boots).

A comparison between bladed boots and soft ground studded boots, as astroturfs do not have suitable traction for use during competitive team sports, has shown that bladed boots have greater injury potential on third generation artificial turf (Bentley, Ramanathan, Arnold, Wang, & Abboud, 2011). When running, blades produced significantly lower peak pressures than soft ground studded boots under the medial heel (167.92 vs. 206.98 Kpa), lateral heel (174.19 vs. 213.8 Kpa), and hallux (430 vs. 461.3 Kpa). Blades also produced significantly lower pressure-time integrals under the medial heel (9.85 vs. 13.01 Kpa.s), lateral heel (10.24 vs. 14.7 Kpa.s), and 1st metatarsal head (49.72 vs. 53.21 Kpa). When running, blades invoked higher pressures on the lateral side, with increased peak pressures under the lateral midfoot (235.94 vs. 219.82 Kpa for bladed vs. soft ground) and 4th metatarsal head (283.37 vs. 265.03 for bladed vs. soft ground boots). When performing a 60 ° cut, bladed boots produced significantly lower peak pressures under the medial heel (244.66 vs. 304.62 Kpa), lateral heel (226.54 vs. 296.23 Kpa), and 1st metatarsal head (373.23 vs. 397.67 Kpa), and higher pressure under toes 3-5 (219.97 vs. 204.33 Kpa) compared to soft ground boots. Greater pressures on the lateral side of the foot suggest that bladed boots may make the athlete more susceptible to ankle inversion or ACL injury, as well as fractures to the 4th and 5th metatarsal heads. Lower medial foot pressures and heel loading reduces the likelihood of chronic overuse injuries or stress fractures on the medial side in bladed boots.

So pressure insoles have also helped to establish the injury potential of different footwear during different sporting movements.

Only two studies have looked at the effect of third generation artificial turf and natural grass surfaces on plantar pressures during dynamic sporting movements. Ford, Manson, Evans et al. (2006) recorded significantly higher peak pressures under the central forefoot and lesser toes when cutting on artificial turf. Ford et al. (2006) fail to report many of their values, but peak pressure was increased by 17.5 % under the central forefoot when cutting on artificial turf, and by 18.9 % under the lesser toes. Significant differences in loading pattern were also recorded, with natural grass producing significantly higher relative loads under the medial forefoot (27.2 vs. 30.2 % for artificial turf vs. natural grass) and lateral midfoot (3.4 vs. 4.1 % for artificial turf vs. natural grass). Ford et al. (2006) hypothesise that the foot is more inverted on artificial turf during cutting, although this is not conclusive from the results provided.

Low (2010) compared plantar pressures during a 180 ° turn and run between artificial turf and natural grass in March and May. In March the artificial turf pitch was harder, as measured by Clegg hammer (93.7 vs. 80 G). During the March test, higher peak pressures were recorded under the medial heel for grass in hard ground boots (21.32 vs. 28.76 $N.cm^2$, artificial turf vs. natural grass) and soft ground boots (18.79 vs. 29.2 $N.cm^2$, artificial turf vs. natural grass). Higher peak pressures were also found on natural grass versus artificial turf for the lateral heel (24.43 vs. 16.06 $N.cm^2$ for hard ground, 28.31 vs. 16.72 $N.cm^2$ for soft ground boots) and 5^{th} metatarsal head (21.21 vs. 17.95 $N.cm^2$ for hard ground, 25.17 vs. 20.77 $N.cm^2$ for soft ground boots). Higher loading rates were also found on natural grass versus artificial turf under the medial heel (.75

vs. .72 N.cm².ms for hard ground, 1.2 vs. .78 N.cm².ms for soft ground boots), lateral heel (.75 vs. .53 N.cm².ms for hard ground, 1.52 vs. .61 N.cm².ms for soft ground boots) and 5th metatarsal head (.87 vs. .62 N.cm².ms for hard ground, .67 vs. .65 N.cm².ms for soft ground boots). This increase in peak pressures and loading rates on the lateral side corresponded with an increase in ankle inversion on grass (24.1 ° vs. 15.1 °, natural grass vs. artificial turf).

In May, the natural grass pitch was harder, as recorded by Clegg hammer (102 vs. 96 G), and recorded significantly higher peak pressures under the medial heel (39.71 vs. 52.66 N.cm² for hard ground artificial turf vs. natural grass, 34.7 vs. 35.94 N.cm² for soft ground boots artificial turf vs. natural grass) and 5th metatarsal head (20.56 vs. 29.15 N.cm² for hard ground artificial turf vs. natural grass, 26.15 vs. 33 N.cm² for soft ground boots artificial turf vs. natural grass). Significantly higher pressure loading rates were also recorded on natural grass versus artificial turf under the medial heel (2.46 vs. 1.58 N.cm².ms for hard ground, 7.88 vs. 1.53 N.cm².ms for soft ground boots) and 5th metatarsal head (1.3 vs. .75 N.cm².ms for hard ground, 1.47 vs. 1.14 N.cm².ms for soft ground boots). Ankle inversion values were slightly higher on natural grass in May (23.7 vs. 21.2 °), but not significantly so.

Aim & Hypothesis

Previous studies have used pressure insoles to assess injury risk during different athletic tasks, on different surfaces, and using different footwear. However there appears to have been a lack of reporting on important injury-related parameters, such as force-time, relative load and load rate variables. Given the importance of establishing injury risk during more dynamic sporting movements, the aforementioned lack of human study on artificial turf surfaces, and the conflicting differences in injury types between artificial turf and natural grass, a study is needed to accurately quantify the injury risk of artificial turf and natural grass during dynamic team sport movements. Therefore the aim of this study is to assess differences in time and injury-related kinetic parameters during two cutting and one running movement, between third-generation artificial turf and natural grass. A movement effect is hypothesised, whereby cutting movements will increase loading on the medial side of the foot. It is also hypothesised that artificial turf will invoke a lateral shift in foot pressures, in line with a more inverted foot position.

Method

Participants

Eight male University rugby players (age 21.13 ± 2.17 y, stature 1.79 ± 0.05 m, weight 86.36 ± 9.6 kg) were recruited for the study. A Cohen's sample size calculation suggested that, for a repeated measures design with two surface and three movement variables, at least six participants were required to produce a significant result, if one was to be found. This was based on a power value of 0.8, *p* value of 0.05 and an effect size of 0.5. Participants comprised of a mix of forwards and backs, and had regular experience of playing and training on both natural grass and third generation artificial turf surfaces. All participants received written and verbal briefs on the test protocol (for Participant Information Sheet, see Appendix A) before signing a written informed consent form (See Appendix B). Participants were able to take part in the study providing they were free from lower body injury within the last 12 months, and had a foot size between UK size 9 ½ and 10 ½. Ethical approval was granted by the University prior to participant recruitment.

Protocol

Each participant attended an habituation session prior to the data collection session. This allowed participants to familiarise themselves with the equipment, movement protocols, and the two test surfaces. This was then followed by the data collection session. Testing was conducted outdoors on an RFU/FIFA approved third generation artificial turf pitch (65 mm pile length, rubber crumb infill), and on an area of natural grass (60 mm pile length). Testing was conducted over three consecutive days, to avoid fluctuations in ground moisture content or hardness, which may have affected the hardness or traction of the sports surfaces, and subsequently kinematic and kinetic responses. Temperature and humidity recorded on the three days of testing were: day one - temperature 19.8 °C, humidity 33 %, day two – temperature 21.5 °C, humidity 34 %, day three – temperature 22.3 °C, humidity 32 %.

The test protocol was set up as a 5 m running channel, with 3 marked lines dictating the 3 different movement protocols (See Figure 1). A straight run, 45 ° cut and a 90 ° cut were chosen as the three movements to assess between the artificial turf and natural grass surface. These movements are distinctly different from each other, and are commonly used in the literature to assess injury risk (Stefanyshyn, Lee & Park, 2010; Muller, Sterzing, Lake & Milani, 2010; Vanrenterghem, Gormley, Robinson & Lees, 2010; Wong et al., 2012).

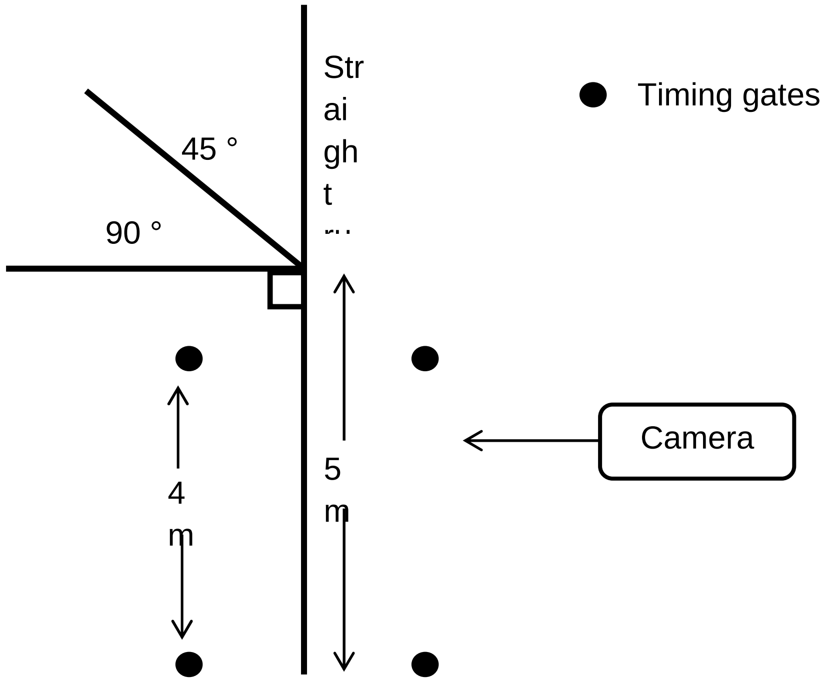


Figure 1: Experimental setup for both the artificial turf and natural grass protocol.

Each participant was required to run straight along the 5 m coned channel, timed at between 3 and 4 m/s, using timing gates along the first 4 m of the channel. In an analysis of task achievement and ACL injury risk during cutting in women, Vanrenterghem, et al. (2012) established that cutting at 3 and 4 m/s allowed for adequate task achievement and lower ACL injury risk. Without a similar study being conducted on males, these values were chosen as the boundaries for acceptable running speed.

On reaching the end of the 5 m channel, participants were required to complete three of each movement protocol. These movements were anticipated so as to increase task achievement. Both the 45 ° and 90 ° cuts were made off the right foot, with the participant instructed to plant their foot as close to the coned line as possible, decelerating and pushing off in one step. For the straight run, participants were asked to continue running to the end of the channel. The first right foot step after the timing gates was used for analysis, including all data from foot contact to toe off. A video camera (Legria FS2000, Canon Inc, Japan, 50 Hz) was placed perpendicular to the 5 m channel to aid identification of the correct right foot step. Participants were given sufficient rest between each trial, to ensure fatigue did not affect movement kinematics and kinetics. Sufficient practice was also given, so that participants completed each movement within their natural gait pattern. Trials were not included if the participant or researcher felt that the designated coned line had been unnaturally targeted.

This protocol was repeated on both artificial turf and natural grass (18 movements in total), with each participant completing the protocol on the same day. Participants wore bladed boots (Adidas 11Questra TRX-FG) which are compliant with both artificial and natural grass surfaces. Into each boot a pressure insole was inserted (Pedar, Novell, USA). Each insole contained 99 sensors, was sampled at 100 Hz, connected to a belt pack, and wirelessly transmitted all data to a laptop. After completion of the movement protocol on one surface, the pressure insoles were removed for 10 minutes, as pressure insole values have been shown to drift after 10 minutes of static loading (Hurkmans, Bussman, Benda, Verhaar & Stam, 2006).

On the first day of testing, a coefficient of restitution ball bounce test was conducted, through the recording of a football inflated to 12 psi, dropped from a height of 1 m, three separate times on each surface. Whilst the coefficient of restitution is not a measure of hardness so much as the interaction between two objects, it gives a crude indication of the difference in elastic or plastic properties of the sports surface.

Statistical Analysis

Analysis of pressure insoles data was conducted using Pedar x.e. (Expert) software. An eight-region mask was used to divide the right foot into sections (See Figure 2). The mask consisted of the Medial Heel (1), Lateral Heel (2), Midfoot (3), Medial forefoot (4), Central Forefoot (5), Lateral Forefoot (6), Hallux (7) and Lesser Toes (8).

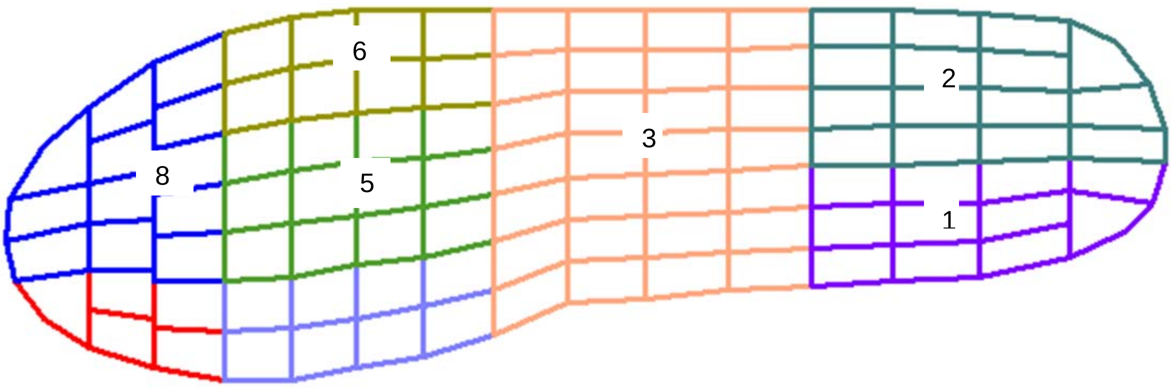


Figure 2: Right foot eight-region mask used for analysis of pressure insole data.

Each region was analysed, for each movement protocol and on each surface, for peak pressure, peak force, pressure-time integral and relative load. Peak force was converted into multiples of body weight. Pressure-time integral was calculated as the area under the pressure-time curve, by the summation of each data point divided by its corresponding time. Relative load was calculated using the force-time integral which was calculated from the peak force data in the same manner as the pressure-time integral. Relative load was then calculated as the percentage of the regional force-time integral to the total foot force-time integral, for each movement on each surface. The total foot was also analysed for contact area and loading rate, for each movement and on each surface. Contact area was calculated as a percentage of the total insole sensor area (NICA). Loading rate was calculated as the peak force value divided by the time taken to reach the peak. Quintic (v.21) was used to establish the coefficient of restitution of each surface-football interaction.

The average of the three trials for each movement protocol was calculated for each individual. Data were parametric and normally distributed, and thus nine 2-way repeated measures ANOVAs were completed using IBM SPSS Statistics 20 for each dependent variable. Each ANOVA represented a region of the foot (with one representing total foot), as comparisons between regions were not needed. The effect of the three movements and two surface conditions on the aforementioned dependant variables was assessed. Paired t-tests were used as post-hoc tests, as surface and movement effects were found, and as such a Bonferroni adjustment to the *p*-value was made, to .017. No statistical tests were conducted on coefficient of restitution data.

Results

A higher average coefficient of restitution was recorded for the third generation artificial turf surface (0.64) compared to the natural grass surface (0.60). Similar speeds were recorded on average for both artificial turf and natural grass protocols, with speeds of 3.31 ± 0.27, 3.32 ± 0.27 and 3.28 ± 0.15 m/s for the straight run, 45 ° and 90 ° cut on artificial turf respectively, compared to 3.3 ± 0.26, 3.25 ± 0.19 and 3.23 ± 0.19 m/s for the straight run, 45° and 90 ° cuts on natural grass.

For the total foot, a significant movement effect was found for the contact area (f=5.062, p <.05). Post-hoc tests revealed a significantly higher total foot contact area on natural grass for the straight run (54.88 %), compared to the 45 ° (48.27 %) and 90 ° (48.73 %) cuts (t=4.202, p<.017, t=3.112, p<.017 for the straight run vs. 45° and straight run vs. 90° cuts respectively). No surface effect was found for the total foot contact area (See Figure 3).

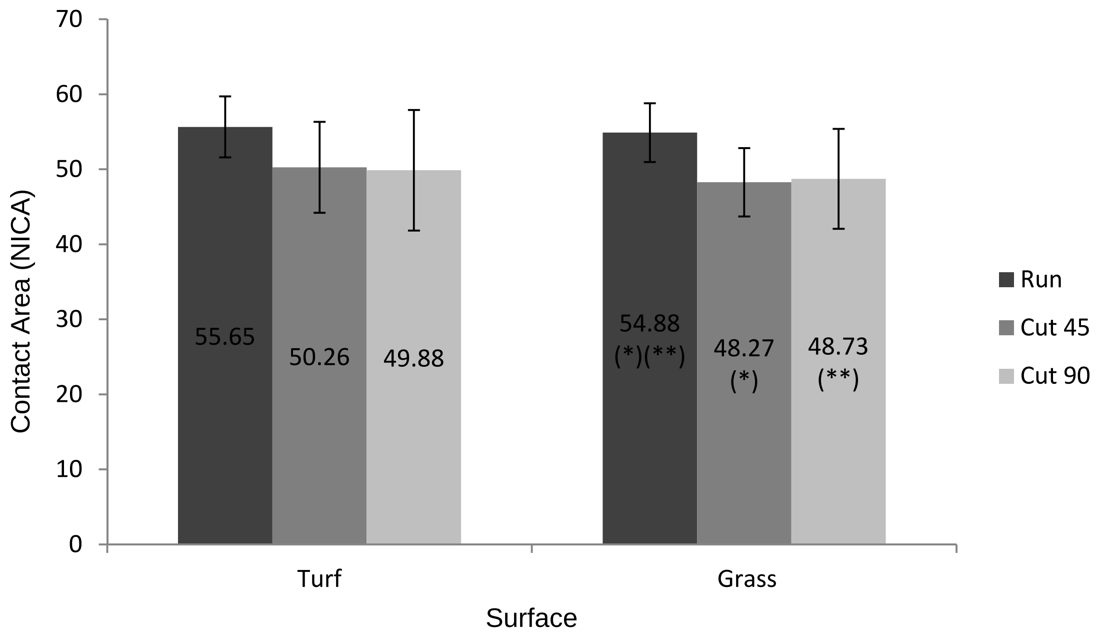


Figure 3: Relative mean contact area of the three movements on artificial turf and natural grass
(*) denotes a significant difference between straight run and 45 ° cut.
(**) denotes a significant difference between straight run and 90 ° cut.

A significant movement effect (f=8.902, p<.05) and surface effect (f=11.843, p<.05) was found for the total foot loading rate. Post-hoc tests showed that only the natural grass straight run recorded a significantly different total foot loading rate than the artificial turf surface (60.22 vs. 39.77 BW/s for natural grass vs. artificial turf, t=-4.277, p<.017). Further post-hoc tests showed that, on artificial turf only, the straight run recorded significantly lower loading rates than the 45 ° cut (39.77 vs. 71.88 BW/s, t=-4.253, p<.017) and 90 ° cut (39.77 vs. 67.94 BW/s, t=-3.589, p<.017) (See Figure 4).

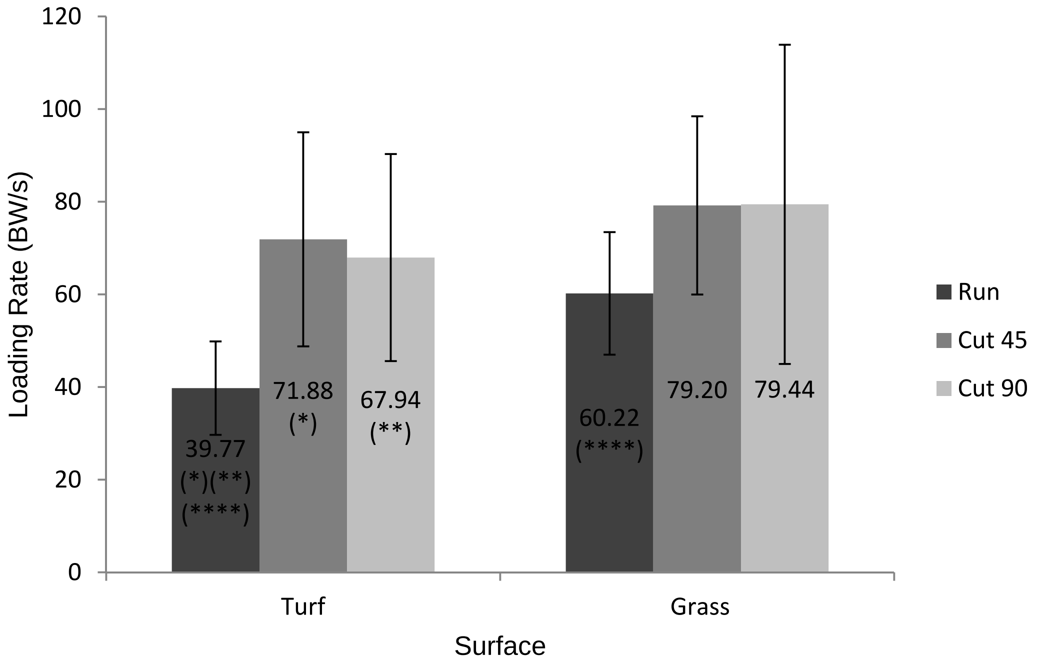


Figure 4: Mean loading rates for the three movements on artificial turf and natural grass.
(*) denotes a significant difference between straight run and 45 ° cut.
(**) denotes a significant difference between straight run and 90 ° cut.
(****) denotes a significant difference between artificial turf and natural grass.

Multiple 2-way repeated measures ANOVAs and post-hoc paired samples t-tests were conducted to assess the effects of movement and surface on peak pressure, peak force and pressure-time integrals over the eight regions of the foot. Significant movement effects were found for all three variables under different regions of the foot, whilst significant surface effects were also found for peak force and peak pressure variables. The results of these can be seen in Tables 1 through 8. Due to the large amount of differences recorded, all relevant SPSS output data, containing *p*-values, f, and t scores, has been included in Appendix C (CD).

Tables 1 & 2: Mean values of peak pressure, force and pressure-time integral, over movement and surface conditions, for the Hallux and Lesser Toes.

Hallux	Run	Cut 45	Cut 90
Turf			
Peak Pressure (Kpa)	433.43 ±67.11	533.25 ± 97.19	470.94 ± 121.48
Peak Force (BW)	.28 ± .06 (*)	.37 ± .08 (*)(***)(****)	.32 ± .08 (***)
Pressure-Time Integral (Kpa.s)	60.11 ± 9 (*)(**)	92.12 ± 27.39 (*)	88.89 ± 27.8 (**)
Grass			
Peak Pressure (Kpa)	429.68 ± 88.46	518.43 ± 67.51	452.18 ± 88.17
Peak Force (BW)	.27 ± .05	.33 ± .07 (****)	.3 ± .06
Pressure-Time Integral (Kpa.s)	57.38 ± 7.91 (*)(**)	89.45 ± 17.57 (*)	88.08 ± 26.42 (**)

Lesser Toes	Run	Cut 45	Cut 90
Turf			
Peak Pressure (Kpa)	212.92 ± 42.17	302.18 ± 85.86	278.33 ± 87.66
Peak Force (BW)	.28 ± .08 (*)	.38 ± .11 (*) (***)	.32 ± .11 (***)
Pressure-Time Integral (Kpa.s)	29.7 ± 6.31 (*)(**)	92.12 ± 27.39 (*)	46.22 ± 13.03 (**)
Grass			
Peak Pressure (Kpa)	205.62 ± 35.05 (*)	314.26 ± 55.79 (*)	285.97 ± 85.05
Peak Force (BW)	.29 ± .07	.38 ± .1	.34 ± .11
Pressure-Time Integral (Kpa.s)	29.69 ± 5.67 (*)	49.12 ± 12.31 (*)	50.33 ± 20.85

(*) denotes a significant difference between straight run and 45 ° cut.
(***) denotes a significant difference between 45 ° and 90 ° cuts.

(**) denotes a significant difference between straight run and 90 ° cut.
(****) denotes a significant difference between artificial turf and natural grass.

Tables 3 & 4: Mean values of peak pressure, force and pressure-time integral, over movement and surface conditions, for the Medial and Central Forefoot.

Medial Forefoot	Run	Cut 45	Cut 90
Turf			
Peak Pressure (Kpa)	357.75 ± 125.87 (*)	477.08 ± 122.28 (*)	458.85 ± 156.03
Peak Force (BW)	.39 ± .14 (*)(**)	.47 ± .16 (*)	.43 ± .17 (**)
Pressure-Time Integral (Kpa.s)	50.33 ± 16.74	83.11 ± 23.44	84.44 ± 32.25
Grass			
Peak Pressure (Kpa)	378.53 ± 123.82	463.54 ± 112.56	442.25 ± 112.57
Peak Force (BW)	.43 ± .15 (*)(**)	.46 ± .2 (*)	.43 ± .17 (**)
Pressure-Time Integral (Kpa.s)	54.23 ± 18.79	83.52 ± 20.83	86.12 ± 25.25

Central Forefoot	Run	Cut 45	Cut 90
Turf			
Peak Pressure (Kpa)	293.85 ± 64.82 (****)	355.93 ± 72.88	320.1 ± 108.83
Peak Force (BW)	.52 ± .11 (**)	.48 ± .09 (***)	.39 ± .09 (**)(***)
Pressure-Time Integral (Kpa.s)	39.42 ± 6.69 (*)	55.23 ± 11.12 (*)	53.89 ± 17.84
Grass			
Peak Pressure (Kpa)	314.68 ± 66.96 (****)	399.06 ± 93.14	322.81 ± 83.68
Peak Force (BW)	.52 ± .12	.48 ± .09 (***)	.39 ± .08 (***)
Pressure-Time Integral (Kpa.s)	43.2 ± 6.99 (*)	64.58 ± 20.14 (*)	58.95 ± 19.99

(*) denotes a significant difference between straight run and 45 ° cut.
(**) denotes a significant difference between straight run and 90 ° cut.
(***) denotes a significant difference between 45 ° and 90 ° cuts.
(****) denotes a significant difference between artificial turf and natural grass.

Tables 5 & 6: Mean values of peak pressure, force and pressure-time integral, over movement and surface conditions, for the Lateral Forefoot and Midfoot.

Lateral Forefoot	Run	Cut 45	Cut 90
Turf			
Peak Pressure (Kpa)	258.64 ± 70.33 (*)(**)	204.68 ± 57.58 (*)(***)	149.89 ± 44.16 (**)(***)
Peak Force (BW)	.32 ± .07 (*)(**)	.25 ± .08 (*)	.19 ± .06 (**)
Pressure-Time Integral (Kpa.s)	35.37 ± 8.32 (**)	30.27 ± 8.19 (***)	25.5 ± 7.05 (**)(***)
Grass			
Peak Pressure (Kpa)	284.05 ± 99.97 (*)(**)	198.33 ± 62.97 (*)	169.78 ± 70.29 (**)
Peak Force (BW)	.36 ± .11(*)(**)	.25 ± .08 (*)	.21 ± .08 (**)
Pressure-Time Integral (Kpa.s)	38.42 ± 11.17	28.99 ± 9.19	28.92 ± 11.81

Midfoot	Run	Cut 45	Cut 90
Turf			
Peak Pressure (Kpa)	187.75 ± 43.52	155.72 ± 25.09	144.68 ± 27.69
Peak Force (BW)	.49 ± .05 (**)	.38 ± .11	.37 ± .1 (**)
Pressure-Time Integral (Kpa.s)	25.31 ± 6.21	20.06 ± 5.75	20.43 ± 6
Grass			
Peak Pressure (Kpa)	183.22 ± 42.15	161.87 ± 29.43	152.18 ± 34.13
Peak Force (BW)	.48 ± .07	.39 ± .11	.38 ± .11
Pressure-Time Integral (Kpa.s)	23.53 ± 5.81	21.29 ± 4.75	22.38 ± 5.79

(*) denotes a significant difference between straight run and 45 ° cut.
(***) denotes a significant difference between 45 ° and 90 ° cuts.
(**) denotes a significant difference between straight run and 90 ° cut.
(****) denotes a significant difference between artificial turf and natural grass.

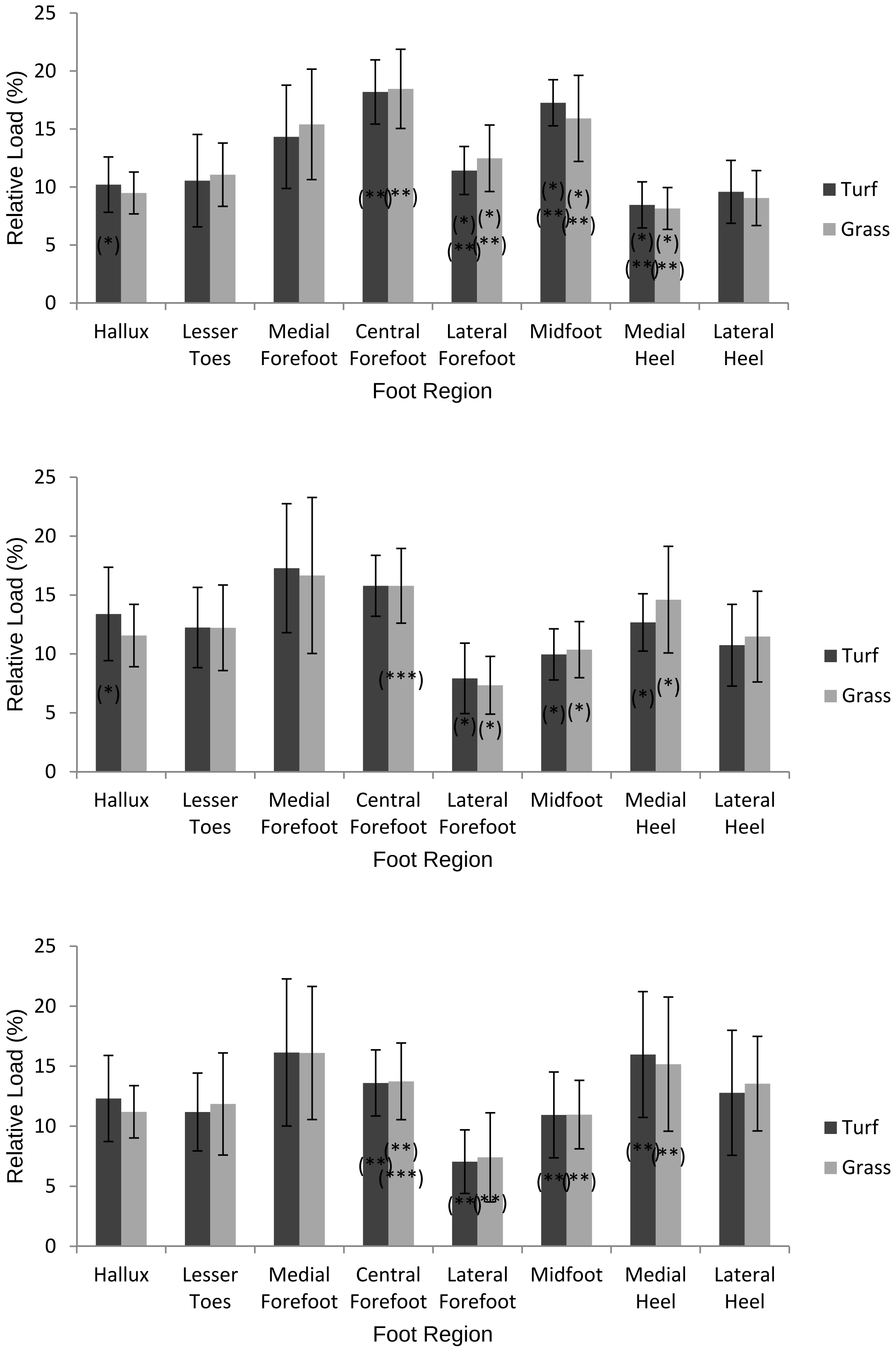
Tables 7 & 8: Mean values of peak pressure, force and pressure-time integral, over movement and surface conditions, for the Medial and Lateral Heel.

Medial Heel	Run	Cut 45	Cut 90
		Turf	
Peak Pressure (Kpa)	248.69 ± 78.56 (*)(**)	496.14 ± 137.42 (*)	487.49 ± 137.73 (**)
Peak Force (BW)	.42 ± .12 (*)(**)	.84 ± .17 (*)	.82 ± .25 (**)
Pressure-Time Integral (Kpa.s)	19.41 ± 6.45 (*)(**)	35.88 ± 10.58 (*)	46.55 ± 16.02 (**)
		Grass	
Peak Pressure (Kpa)	273.33 ± 68.76 (*)(**)	560.41 ± 97.76 (*)	532.91 ± 146.84 (**)
Peak Force (BW)	.45 ± .09 (*)	.89 ± .19 (*)	.82 ± .33
Pressure-Time Integral (Kpa.s)	19.5 ± 5.34 (*)(**)	45.52 ± 14.99 (*)	51.99 ± 18.95 (**)

Lateral Heel	Run	Cut 45	Cut 90
		Turf	
Peak Pressure (Kpa)	245.72 ± 79.59(*)(**)	459.97 ± 128.69 (*)	446.45 ± 122.15 (**)
Peak Force (BW)	.48 ± .14 (*)(**)	.73 ± .16 (*)	.69 ± .17 (**)
Pressure-Time Integral (Kpa.s)	18.82 ± 6.11(*)(**)	30.77 ± 8.78 (*)	38.06 ± 13.91 (**)
		Grass	
Peak Pressure (Kpa)	263.37 ± 71.78 (*)(**)	518.33 ± 96.44 (*)	470.62 ± 118.18 (**)
Peak Force (BW)	.51 ± .13 (*)	.83 ± .14 (*)	.76 ± .19
Pressure-Time Integral (Kpa.s)	18.06 ± 5.01 (*)(**)	35.55 ± 12.52 (*)	42.53 ± 14.05 (**)

(*) denotes a significant difference between straight run and 45 ° cut.
(***) denotes a significant difference between 45 ° and 90 ° cuts.
(**) denotes a significant difference between straight run and 90 ° cut.
(****) denotes a significant difference between artificial turf and natural grass.

A significant movement effect was found for relative load under the Medial Heel ($f=10.807$, $p<.05$), Midfoot ($f=30.597$, $p<.0005$), Lateral Forefoot ($f=20.782$, $p<.0005$), Central Forefoot ($f=11.473$, $p<.05$) and Hallux ($f=6.223$, $p<.05$). No surface effect was found for relative load under any foot region. Post-hoc paired t-tests were conducted to establish movement effects on artificial turf and natural grass, and can be found in Figures 5-7. Again, the large amount of differences prevents the reporting of all statistical test values, however they can be found in Appendix C (CD).



Figures 5-7: Mean relative load under each foot region for straight run (top), 45 ° (middle) and 90 ° (bottom) cuts on artificial turf and natural grass.
(*) denotes a significant difference between straight run and 45 ° cut.
(**) denotes a significant difference between straight run and 90 ° cut.
(***) denotes a significant difference between 45 ° and 90 ° cuts.

Discussion

A coefficient of restitution ball bounce test showed that, whilst the artificial turf had a slightly more elastic response than the natural grass surface, values were very similar (0.64 vs. 0.6). Although this is not a direct measure of hardness as much as a measure of the elasticity or plasticity of an interaction between two objects, it is very much affected by the hardness of the surface. Therefore it can be assumed that the two surfaces were similar in their levels of hardness at this time of year (June). Similar running speeds were also recorded for all movements on both surfaces, with the maximum difference between movements and surfaces 0.09 m/s. It can therefore be assumed that differences observed in kinetic and kinematic variables are not due to inconsistencies in running speed.

The two cutting manoeuvers were found to illicit lower total foot contact areas than the straight running on both surfaces, although this was only significant on grass. The greater variation in foot contact area values on artificial turf may be the cause of the lack of statistical significance. A difference between artificial turf and natural grass of no more than 2 % of the total foot area within each movement category suggests little difference in movement patterns between the surfaces. However a reduction in total foot contact area for cutting movements compared to running highlights a change in movement patterns. A reduction in contact area will also increase plantar pressure values, if force values stay uniform throughout each movement. Queen et al. (2008) recorded total foot contact area during the cut of .903 NICA, much higher than the values recorded in this study (48.27 – 50.26 NICA). It is not clear how Queen et al. (2008) calculated the total foot contact area, but the higher value suggests that contact area during the cut was calculated as a percentage of the maximum area recorded, as opposed to the total area of the pressure insole.

Whilst contact area under different foot regions has been assessed in other studies (Putti et al., 2007; Queen et al., 2010; Wang et al., 2012), the errors associated with subsampling of pressure sensor areas (Pataky & Maiwald, 2011) are likely to make these results incomparable. Therefore total foot contact area was chosen as a more accurate measure of loading area.

In terms of loading rate, grass was found to have a significantly higher loading rate for all three movements, compared to artificial turf. However this was only significant for the straight run. Again the large variation in loading rate values may be the cause of the lack of between-surface significant differences for the two cutting movements. The greater loading rates on grass would suggest that the grass surface was slightly harder, as harder surfaces are generally illicit higher loading rates (Stiles & Dixon, 2007). High loading rates are also associated with increased risk of overuse injuries, through exposure of the bodies shock absorbing mechanisms to high levels of force. The highly variable nature of natural grass surfaces however, which are mostly dictated by seasonal environmental conditions, means that hardness properties and subsequently loading rate responses will fluctuate throughout the season. Furthermore the time of testing was outside of the rugby and football playing season, so it is not the case that grass increases the likelihood of overuse injury throughout the playing season. It can only be concluded that on this natural grass pitch, the likelihood of overuse injury is higher than that of artificial turf, at this time of year.

Dixon et al. (2000) recorded similar values for loading rates on asphalt (42.9 – 62.3 BW/s) and acrylic (42.1 - 56.7 BW/s) during running, compared to this study (39.77-60.22 BW/s). It would be expected that harder surfaces such as asphalt would illicit higher loading rates than grass or artificial turf surfaces. However the greater shock absorbing properties of running shoes compared to rugby boots, as well as the pronounced heel that works to increase contact time and reduce loading rate, are probably responsible for the similarity in loading rate values here. Stiles et al. (2011) recorded higher loading rates on clay (84.67 BW/s) and sand-based (96.74 BW/s) grass pitches during running than in this study. However the difference in running speeds (3.32 vs. 3.83 m/s) may be the cause of the difference in loading rates, with greater running speeds invoking greater ground reaction forces and loading rates.

Very few between-surface differences were found in this study, contrary to the findings of Ford et al. (2006). Hallux peak force was found to be significantly higher on artificial turf than natural grass during the 45 ° cut (.37 vs. .33 BW). A higher, but not significantly so, relative load under the Hallux on artificial turf during the 45 ° cut (13.39 vs. 11.56 %) may be indicative of a greater shift towards the Hallux during cutting on turf, and may explain the higher force values found here. Higher peak pressure was also recorded under the central forefoot during running on grass compared to artificial turf (314.68 vs. 293.85 Kpa). Similar force values were recorded between turf and grass under the central forefoot during all movements, yet turf consistently produced lower peak pressures under this region. Therefore differences in pressure must be due to a change in contact area. The aforementioned error associated with comparing contact area using pressure insoles prevented a comparison between regions in this study, so this hypothesis cannot be confirmed. No difference in relative load under the central forefoot however suggests that running on grass increases the likelihood of stress fractures under the 2nd and 3rd metatarsal areas that this region represents. How much greater a risk this difference represents is unclear, as the relationship between plantar pressures and metatarsal fractures has not been established.

No other between-surface differences were recorded in this study. This would suggest that both surfaces respond similarly, and there is no change in loading pattern that may be associated with an increase in ACL or ankle sprain injuries on either surface. Ford et al. (2006) recorded a greater medial shift in plantar loading when cutting on grass, meaning artificial turf allowed for greater foot inversion. It is this movement that is associated with both ACL and ankle inversion sprains; however this study contradicts the findings of Ford et al. (2006), in that no loading differences can be found that may relate to differences in ACL or ankle inversion sprain injury risk. It may be that seasonal differences in time of testing and grass surface conditions may affect loading patterns. Ford et al. (2006) reported the artificial turf surface as harder than grass, whereas the levels of hardness in this study were much closer, with grass the slightly harder surface. Repeated testing over the course of a season would better assess ACL and ankle inversion injury risk throughout different grass surface conditions.

Whilst surface has not been found to affect ACL or ankle inversion injury risk in this study, movement type has been shown to affect the likelihood of such injuries. On artificial turf, significantly higher pressures, and pressure –time integrals were recorded under the Hallux, Lesser toes, Medial Heel, Lateral Heel and Central Forefoot for both cutting movements compared to running.

This resulted in higher relative loads under the Hallux, Central Forefoot and Medial Heel higher during cutting compared to running on artificial turf, which is indicative of a medial shift in plantar loading. Peak force follows the same pattern between running and cutting, except under the central forefoot, which recorded higher values during running on turf.

Between cutting movements, significantly higher peak forces were recorded under the Hallux, Toes and Central Forefoot for the 45 ° cut. Pressure and pressure-time integral values were not different under these foot regions, so differences in force values between cuts suggests that greater force production occurs under the forefoot during the 45 ° cut compared to the 90 ° cut. The 90 ° cut is a lateral propulsion of the body, so it stands to reason that the 45 ° cut, which has a more forward propulsion of the body, produces greater ground reaction forces under the forefoot, where propulsive forces are generated against the ground.

The lateral forefoot and midfoot recorded the lowest peak pressure, peak force and pressure-time integral values during the 90 ° cut, followed by the 45 ° cut, then the straight run. The decrease in total foot contact area during cutting movements may be explained by the medial shift in loading during cuts, and the reduced relative load on the lateral portion of the mid and forefoot. However despite a shift in loading to the medial portion of the foot during cuts, no difference in Medial Forefoot relative load was recorded. This was also despite an increase in peak pressures, force and pressure-time integrals under this region of the foot during cuts compared to straight running. Loading at the heel was much higher at 45 ° and 90 ° than straight running, at values proportionately higher than the increase seen under the Medial Forefoot.

Therefore the increased loading at the Medial Forefoot may have been masked by the disproportionate increase in loading at the heel. It should be noted therefore that relative load is not as sensitive to changes in loading patterns as previously suggested, and should be assessed alongside other values of loading, such a peak pressure and pressure-time integrals.

Higher Medial heel relative load was recorded for 90 ° compared to 45 ° cut, although this was not significant (15.97 vs. 12.68 %). This higher loading at the heel would support the notion that the 90 ° cut requires a more rapid deceleration and less emphasis on the forefoot propulsion, so explains the differences in peak pressures and force under the Hallux, Lesser Toes and Central Forefoot between the two cuts. The higher medial foot plantar loading during cutting movements suggests the risk of sustaining an ACL and ankle inversion sprain injury on artificial turf through rapid foot inversion is low. However there are other kinematic factors associated with development of ACL injury. Higher medial loading is indicative of high loading on the medial knee compartment. As previously mentioned, slackening of the lateral knee ligaments through compression of the medial knee compartment can increase the likelihood of anterior Tibial displacement and subsequently ACL injury. So it may be the case that ACL injuries occur through means other than foot inversion during cutting. Higher peak pressures and pressure-time integrals under the Lateral Forefoot during running compared to cutting on artificial turf may be suggestive of lateral metatarsal fractures. Whilst the 1st and 2nd metatarsal heads are built to withstand high levels of force, the lateral metatarsals are more susceptible to fractures under high loading. However it is unclear what level of pressure, or pressure loading, the lateral metatarsals are able to withstand, so it is unclear how much of an increased risk of fractures running imposes on the lateral metatarsal heads.

The medial shift in loading under the foot during cutting has been reported in other studies. Eils et al. (2004) reported a comparable shift in Hallux relative load from running to a cut (9.6 to 13.3 %, compared to 10.21 to 13.39 % in this study). Similar increases in relative load were recorded in the Medial Forefoot (18.7 to 25.8 %, compared to 14.32 to 17.27 % in this study) and Medial Heel (7.8 to 15.9 %, compared to 8.45 to 12.68 % in this study) during cuts compared to running, with values slightly higher than those recorded in this study. The slightly higher loading at the medial heel and forefoot may be due to the differences in running speed (4.2 vs. 3.3 m/s on average), although Eils et al. (2004) do not specify their cutting angle, which if different may invoke slightly different kinematic and kinetic responses. Whilst Eils et al. (2004) recorded a drop in relative load under the Lateral Forefoot (18.2 to 5.7 %) and Central Forefoot (15.6 to 10.5 %) during cuts compared to running, these values were higher than those recorded in this study (11.42 to 7.9 % for the Lateral Forefoot and 18.19 to 15.78 % for the Central Forefoot). Whilst the loading patterns remain the same and results between the two studies are still very comparable, with no more than 5 % difference in loading between each region, the participants used in this study held a more neutral foot position than those used by Eils et al. (2004). This would explain the slightly lower medial and slightly higher lateral loading under the foot during running and cutting.

Wong et al. (2006) also found greater medial loading under the foot during cutting compared to running. In particular, greater peak pressures under the Hallux, Medial Forefoot and Central Forefoot were found for cutting, as was found in this study. A reduction in pressure under the Lateral Forefoot was also recorded by Wong et al. (2006). Compared to this study, higher pressures under the Medial Forefoot (550 Kpa), as well as lower pressures under the Central Forefoot (347 Kpa) and Lateral Forefoot (79 Kpa) during cutting, highlight the more neutral foot position of participants recruited in this study (477 Kpa Medial Forefoot, 355 Kpa Central Forefoot, 204 Kpa Lateral Forefoot). But despite this, uniform changes in loading patterns between running and cutting are evident with this study and the literature.

Low (2010), contrary to the results of this study, reported higher pressures under the 5^{th} metatarsal than the 1^{st} metatarsal head during a turning movement (20.56 N/cm^2 vs. 16.75 N/cm^2). A 180 ° turn, as was used as the movement protocol by Low (2010) clearly invokes different kinematic and kinetic responses than cutting, and is therefore not comparable to cutting movements. The increased lateral loading does however suggest greater foot inversion. In terms of ankle inversion sprain and ACL injury through inversion mechanisms, 180 ° turns may pose a greater risk. But the lower number of executions of 180 ° turning movements in evasive team sports such as rugby, compared to cutting movements, reduces the importance of assessing plantar loading during this movement.

Fewer significant differences between movements were recorded on the natural grass surface. Higher variations in kinetic and kinematic parameters on grass may be reflective of the less uniform state of the surface, and are probably responsible for the lack of statistically significant differences. Differences between the three movements on grass follow the same patterns as on artificial turf, and therefore it is likely that the uneven state of the grass surface has caused a number of type II errors to be made. The natural grass surface recorded higher peak pressures under the Hallux, Lesser Toes, Medial Forefoot, Central Forefoot, Medial Heel and Lateral Heel for cuts compared to the straight run, although this was only significant for the Medial Heel and lesser Toes. Pressure-time Integral and Peak force followed the same pattern, except under the Central Forefoot where once again a lower peak force was recorded for cuts compared to straight running. Once again, higher peak pressures and pressure loading occurred under the medial portion of the foot, and it is probable that the lower total foot contact area during cutting is a result of reduced contact area under the Lateral Forefoot.

The same pattern in relative load occurred during running and cutting on grass as with artificial turf. Significantly higher relative load was found for the Medial Heel for cutting compared to straight running, and non-significantly higher relative load was found at the Hallux. Significantly lower relative load under the Lateral and Central Forefoot was found for cutting compared to running, although this was non-significant for the 45 ° cut under the Central Forefoot. As was the case with the artificial turf surface, despite an increase in peak pressure, force and pressure loading under the Medial Forefoot during cutting, no significant change in relative load was found. It is predicted that disproportionately high increases in loading at the Medial Heel mask the increased loading at the Medial Forefoot. It is therefore advocated that relative load be assessed alongside other measures of loading at specific foot regions.

In terms of the kinetic values recorded for each movement on natural grass, good conformity was also found with the literature. During running, Tessutti et al. (2010) recorded similar values of peak pressure and pressure-time integral to this study under the Medial Heel (276 vs. 273 Kpa, 19.9 vs. 19.5 Kpa.s), Lateral Heel (283 vs. 263 Kpa, 17.9 vs. 18.06 Kpa/s), Medial Forefoot (337.7 vs. 378.53 Kpa, 45.2 vs. 54.23 Kpa/s) and Lateral Forefoot (214.5 vs. 284.05 Kpa, 30.5 vs. 38.42 Kpa/s). Wang et al. (2012) also recorded similar values for peak pressure and normalised peak force during running on grass, under the Medial Heel (204.9 vs. 273.33 Kpa, 49.6 vs. 45 %BW), Medial Forefoot (345.1 vs. 378.53 Kpa, 43.7 vs. 43 %BW), Central Forefoot (336.8 vs. 314.68 Kpa, 52.1 vs. 52 %BW), Lateral Forefoot (257.9 vs. 284.05 Kpa, 34.1 vs. 36%BW), Hallux (365.2 vs. 429.68 Kpa, 32.1 vs. 27 %BW) and Lesser Toes (212.4 vs. 205.62 Kpa, 33.6 vs. 29 %BW). Finally Hong et al. (2012) recorded peak pressure and force values similar to those in this study, during running on grass, in the Medial Heel (214.2 vs. 273.33 Kpa, 54.2 vs. 45 %BW), Lateral Heel (207 vs. 263 Kpa, 44.8 vs. 51 %BW), Medial Forefoot (331.2 vs. 378.53 Kpa, 46.6 vs. 43 %BW), Central Forefoot (331.5 vs. 314.68 Kpa, 56 vs. 52 %BW), Lateral Forefoot (247.6 vs. 284.05 Kpa, 35.2 vs. 36 %BW) and Lesser Toes (231.8 vs. 205.62 Kpa, 36 vs. 29 %BW). The small differences between values found in other studies are negligible, given that natural grass surfaces are highly variable and less stable in their characteristics than artificial turf. The medial shift in plantar loading during cutting has been confirmed by other studies in the literature, and has been found in this study on both artificial turf and natural grass surfaces, thus fulfilling the hypothesis that cutting will increase loading on the medial portion of the foot. The second hypothesis, that artificial turf would invoke a more lateral loading of the plantar surface, can be rejected.

Ford et al. (2006) conducted the only study to assess cutting movements on artificial turf and natural grass, using appropriate measures of plantar loading such as relative load. Whilst Ford et al. (2006) fail to give values for most of their variables, the highest pressure areas under the foot were the Medial Forefoot and Hallux, as was the case in this study. This study also recorded high pressure values and pressure loading under the Medial and Lateral Heel during cutting. Perhaps the movement protocol, set up as an agility course in the study by Ford et al. (2006), reduced the need for high braking forces, and is the cause of this difference between the two studies. Ford et al. (2006) also recorded a difference in relative load between grass and artificial turf, with greater loading on the Medial Forefoot on grass during cutting indicative of a medial shift in plantar loading. This was not found to be the case in this study, and thus the 2nd hypothesis has been rejected. The reason for differences between these two studies, in terms of the conclusions drawn about grass and artificial turf surfaces, are unclear. Natural grass is a highly variable and easily influenced surface, which constantly changes in measures of hardness and traction throughout the year. Furthermore there are variations in grass types and soil types which can affect lower body kinetics and kinematics (Stiles et al., 2011). It is highly possible that changes in the grass surface are responsible for the differences observed in the study by Ford et al. (2006), whilst in these study; grass reacted very similarly to artificial turf.

Relating differences between artificial and natural grass surfaces to differences in injury types between the surfaces, there is no clear explanation for the differences in ACL and ankle inversion sprains recorded in team sports such as football, American football and rugby. Conflicting injury data are likely to be explained by the high seasonal variation in natural grass surfaces. Higher loading on the medial portion of the foot during cutting contradicts many predictions that ACL and ankle inversion injuries most commonly occur during cutting movements (Kristianslund et al., 2011; Dempsey et al., 2007; Donnelly et al., 2012; Ford et al., 2005). It is possible that the reason for this is the anticipated nature of the movement protocol. Besier et al. (2001) found that unanticipated cutting, compared to anticipated cutting, increased knee Valgus loading 12-fold, and knee internal rotation by 129 %, thus drastically increasing ACL strain and injury risk. Whether this is accompanied by rapid ankle inversion is unclear, however it may be the case that unanticipated cutting does not allow for the correct and safe cutting procedure to be executed, resulting in greater ankle inversion. This is perhaps an area for future studies to assess.

Conclusion

During straight running and anticipated cutting on artificial turf and natural grass, no significant kinetic or kinematic differences exist between surfaces. Whilst higher loading rates were recorded on grass, significantly so for running, the variable nature of the hardness of natural grass surfaces make it highly unlikely that this is a uniform and consistent difference. Furthermore, as testing was conducted outside the rugby playing season, injury-potential related to differences in surface hardness may be moot. The hypothesis that the artificial turf surface would invoke greater lateral foot pressures, in line with the findings of Ford et al. (2006), has been rejected. Contrary to the findings of Ford et al. (2006), no differences in loading patterns, or kinetic measures were observed between the surfaces. Both surfaces responded similarly during straight running and two cutting manoeuvers, and thus the differences in ACL and ankle inversion sprain injuries sustained during team sports between the two surfaces cannot be explained through these findings.

Whilst no surface effect was found in this study, a significant movement effect was observed, accepting the first hypothesis, whereby cutting movements caused a medial shift in loading underneath the foot. It is hypothesised that this medial shift in loading could kinematically predispose players' to anterior Tibial displacement, through compressing the medial knee compartment and slackening the lateral knee ligaments which maintain the position of the Tibia. This would contradict the majority of literature, which states that ACL and ankle inversion sprain injuries occur through rapid inversion of the ankle. This was not seen to be an injury mechanism in this study. It is possible that the anticipated nature of the movement protocol allowed participants to adjust their lower body posture, and correct any movements such as excessive foot inversion, which may be a precursor to ACL and ankle inversion injuries. So this study has found that the type of movement affects injury propensity, more so than whether the surface is artificial or natural grass. The highly variant nature of natural grass surfaces will mean that these results are not uniform, and will vary between time of year and between different grass/soil compositions. Given that, during anticipated dynamic sports movements, no difference in injury potential has been observed between surfaces, perhaps the benefits of having a uniform and consistent artificial turf surface outweigh the aesthetic needs for a natural grass sports surface in dynamic team sports.

This study also highlights the need and use of human testing and field testing in ascertaining the suitability and injury potential of sports surfaces. The unique response of the dynamic human shock absorbing system to different sports surfaces cannot be replicated by machinery. Furthermore, dynamic sports movements such as cutting hold greater injury-potential than running, which is typically used as a benchmark to assess sports surfaces. The testing of sports surfaces in-situ, whilst including environmental constraints and considerations, allows for a normal and accurate response from both the sports surface and interacting body, be it athlete or ball. Pressure insoles provide a reliable and valid tool for field testing, and understanding plantar loading and injury potential during dynamic sporting movements. The calculation of further measures, such as relative load, allow for understanding of foot loading patterns. However it is thought that in this study higher levels of loading at the Medial Heel masked increasing in loading at the Medial Forefoot. Therefore relative load data should be interpreted with caution, preferably alongside other kinetic data such as peak pressure or pressure-time integrals.

This study was limited by the anticipated nature of the movement protocol. Unanticipated cutting has been shown to increase knee valgus and internal rotation, which are both linked to ACL injury. Whether surface-related differences in ACL or ankle inversion sprain injury risk exist during unanticipated cutting needs to be assessed before conclusive judgements can be made about the suitability of natural grass and artificial turf surfaces for dynamic team sports. This study was also limited by the speed at which cutting movements were performed. Whilst speed was limited to increase task achievement, this is unlikely to reflect the speed at which movements such as cutting are performed during team sports. The more rapid change of direction may be a further cause of surface-related injury differences during cutting, although this comes with ethical considerations, as cutting at speeds exceeding 4 m/s have been shown to increase knee valgus to injury-inflicting levels (Vanrenterghem et al., 2012).

Further research should look to assess the effects of increased running speed and unanticipated dynamic sports manoeuvers on plantar loading on artificial turf and natural grass. This may provide us with the most comprehensive analysis of foot-ground interactions in team sports, and surface-specific mechanisms for ACL and ankle inversion injuries. Further research should also focus on techniques to reduce knee valgus loading and ankle inversion loading, in order to reduce the risk of ACL and ankle sprain injuries during dynamic team sports. As no surface-specific differences in loading patterns have been found in this study, perhaps more focus should be given to areas such as adjusting technique to reduce injury risk (Myers & Hawkins, 2010).

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Appendix A – Participant Information Sheet

Participant information sheet

Plantar pressure measures of rugby-specific movements on artificial turf and natural grass.

You are being invited to take part in a research study. Before you decide, it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

Thank you for reading this.

What is the purpose of the study?

This research is being undertaken on healthy adult rugby players, free from lower body injuries in the last 12 months. The purpose of this study is to determine how differences in pressures underneath the foot, between running and different-angled cuts made on artificial turf and natural grass, can explain differences in lower body injuries between the two surfaces. The research mainly aims to look at the effect of the surface on the risk of knee and ankle ligament injuries, as these are often caused by twisting or turning movements.

Why have I been chosen?

You have been chosen because you are a part of the University Rugby Union or League Mens team, and fulfil the equipment requirements (Size 9 ½- 10 ½ shoes). You also have had no ankle, knee, or other lower body injuries in the last 12 months.

Do I have to take part?

It is up to you to decide whether or not to take part. If you decide to take part you will be given this information sheet to keep and be asked to sign a consent form. If you decide to take part you are still free to withdraw at any time and without giving a reason. A decision to withdraw at any time, or a decision not to take part, will not affect you in any way.

What will happen to me if I take part?

If you decide to take part, you will be given this information sheet to keep and asked to sign the consent form. This will give your consent for a researcher from the University of Chester to contact you to invite you to attend a familiarisation session at a time of your convenience. This session is a mock-version of the data collection session, and will allow you to get used to the procedures and equipment that will be used. You will then be contacted to invite you to the data collection session. This session will involve you performing 2 sidesteps, at 45 and 90 °, and a straight run, on both artificial turf and grass. For this data collection session you will be wearing pressure insoles in the shoes provided, and with your permission, the session will be video recorded. This will conclude your involvement in the study.

What are the possible disadvantages and risks of taking part?

A thorough warm up protocol will be conducted prior to data collection, so the risk of injury is low. The study should take 1.5 hours of your time, split into a 30 minute habituation session and a 1 hour data collection session.

What are the possible benefits of taking part?

The study will allow you to experience new Biomechanical equipment and understand how it may be used and applied in sports settings. If you wish you may receive feedback on your loading patterns when cutting on the two surfaces. By taking part you will also be contributing to the understanding of the effects of sports surface on rugby-specific movements.

What if something goes wrong?

If you wish to complain or have any concerns about any aspect of the way you have been approached or treated during the course of this study, please contact Professor Sarah Andrew, Dean of the Faculty of Applied Sciences, University of Chester, Parkgate Road, Chester, CH1 4BJ, 01244 513055.

Will my taking part in the study be kept confidential?

All information which is collected about you during the course of the research will be kept strictly confidential so that only the researcher carrying out the research will have access to such information.

What will happen to the results of the research study?

The results will be written up into a dissertation for my final project of my MSc. Individuals who participate will not be identified in any subsequent report or publication.

Who is organising the research?

The research is conducted as part of a MSc in Sports Sciences (Biomechanics) within the Department of Sport & Exercise Sciences at the University of Chester. The study is organised with supervision from the department, by Matthew Page, an MSc student.

Who may I contact for further information?

If you would like more information about the research before you decide whether or not you would be willing to take part, please contact:

Matthew Page.

Thank you for your interest in this research.

Appendix B – Informed Consent Form

University of Chester

Title of Project: Plantar pressure measures of rugby-specific movements on artificial turf and natural grass.

Name of Researcher: Matthew Page

Please initial box

1. I confirm that I have read and understand the information sheet for the above study and have had the opportunity to ask questions. ☐

2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason and without my legal rights being affected. ☐

3. I agree for the data collection session to be video recorded. ☐

4. I agree to take part in the above study. ☐

____________________	__________________	______________
Name of Participant	Date	Signature
__________________________	________________	_________________
Researcher	Date	Signature